

**A Method to Correct Organ Motion Artifacts in X-ray CT Systems Based on
Tracking of Motion Phase by the Spatial Overlap Correlator**

by

Amar C. Dhanantwari

57 Stillwater Crescent
Toronto, ON M3R 3S3
(416) 663-1813

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on behalf of
DEPARTMENT OF NATIONAL DEFENCE

as represented by
Defence and Civil Institute of Environmental Medicine
1133 Sheppard Avenue West
Toronto, Ontario, Canada M3M 3B9

DCIEM Scientific Authority:
Dr. S. Stergiopoulos
(416) 635-2049

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Executive Summary

This report describes a novel method to track organ motion and eliminate motion artifacts in x-ray Computed Tomography (CT) cardiac imaging applications with emphasis on the detection of cardiac blood vessel calcification. The method consists of a spatial overlap correlator scheme to track organ motion in x-ray CT systems. A coherent sinogram synthesis (CSS) method is then used to correct for organ-motion artifacts. Since the spatial overlap correlator scheme tracks the cardiac motion, it identifies also the projection points associated with minimum amplitude cardiac motion effects. These points are used to identify a 180 degrees plus the fan angle sinogram for image reconstruction. This leads to a retrospective gating (RG) scheme that is based on the output results of the spatial overlap correlator. Performance comparison of the proposed methodology with the retrospective ECG-gating using real data sets with phantoms and human patients provides a performance assessment of the merits of the proposed methods. Real results demonstrate that the new methodology eliminates the requirement for ECG, the CSS method does not require breath-holding and that it is possible to be implemented in x-ray CT scanners to detect cardiac blood vessel calcification and image the heart's ventricles.

I. INTRODUCTION

X-ray Computed Tomography is an imaging modality that provides images of transverse cross sections or "slices" of the human body. The patient is placed in between an X-ray source and an array of sensors; the latter measuring the amount of radiation that passes through the body. During the data acquisition process, source and detectors move around the patient resulting in a large number of X-ray projections. This process assumes stationarity, which is violated when there is organ motion (e.g. cardiac motion, blood flow, lung motion due to respiration, patient restlessness) during the relatively long data acquisition process (e.g. 0.5 to 1.5 seconds). Thus, the reconstructed image reconstruction is adversely affected by the so-called motion artifacts, which appear as blurring and distortion in the reconstructed images, and may lead to inaccurate diagnosis [1,2]. The intuitive solution to this problem is to speed up the data acquisition process, so that the motion effects would become negligible. This requirement, however, cannot be met with current technology. The scan times that are required are simply not realizable even by the most advanced conventional CT scanner today. The only CT scanner that delivers the required speed is the Electron Beam CT (EBCT), which operates at very low signal-to-noise ratio and suffers from poor image resolution. Signal processing algorithms to account for organ motion artifacts are therefore needed.

Several mathematical techniques have been proposed as solutions to this problem. These techniques usually assume a simplistic linear model for the motion, such as translational, rotational or linear expansion [4]. Some techniques model the motion as a periodic sequence and take projections at a particular point in the motion cycle to achieve the effect of scanning a stationary object (similar to ECG triggering) [3]. However, in practical applications, motion is much more complex, and these techniques are useful in a very limited number of cases. A more general technique attempts to iteratively suppress the motion effects from the projection data [5]. This method attempts to reduce the assumed spectral characteristics of the organ motion. It, however, depends on knowing some properties of the motion a-

priori, and requires a number of iterations to converge. The spectral properties of some complex form of motion may not necessarily correspond to the model used in this algorithm. Moreover, without proper initialization, the convergence period could be long.

Motion artifacts in magnetic resonance imaging (MRI) for chest scans have been reduced by first defining the motion with a parametric model and then adapting the reconstruction algorithm to correct for the modeled motion [6]. Ritchie et al [7] attempted to address CT cardiac and respiratory motion artifacts by using a pixel-specific back-projection algorithm that was conceptually influenced by the MRI motion modeling approach [6]. In the CT case, the frame of reference was specified by constructing maps that describe the motion of each pixel in the image at the time each projection was measured [6]. These maps then formed the basis of a back-projection algorithm that reconstructed each pixel in a frame of reference that moved according to the information provided by the maps. However, the method requires substantial manual effort to describe the motion of each pixel and therefore is not practical for x-ray CT medical imaging applications.

The problem of motion artifacts, however, is well known in other type of real time imaging systems such as radar satellites and sonars [8,9]. In this case, it has been found that the application of synthetic aperture processing increases the resolution of a phased array imaging system and corrects for the motion artifacts as well. Reported results [8,9] have shown that the problem of correcting motion artifacts in synthetic aperture sonar applications is centered on the estimation of a phase correction factor. This factor is used to compensate for the phase differences between sequential sensor array measurements to coherently synthesize the multiple spatial measurements into a synthetic aperture. The processing concept of this investigation is based on the spatial overlap correlator algorithm [8,9] that includes cross-correlation of pairs of sensors that overlap in space.

This paper attempts to address the problem of CT motion artifacts with a novel signal processing concept that is implemented at the CT sensor time series (sinograms) before the application of any type of image

reconstruction algorithm. In this effort, no assumptions are made about the type of motion models that would describe the organ-motion. The signal processing structure of this investigation includes three basic steps:

- First, the spatial overlap correlator (SOC) [8,9,10] is implemented in x-ray CT systems as a method of tracking the phases of the organ motion.
- Second, a processing step entitled "coherent sonogram synthesis" (CSS) that identifies through a replica correlation process the segments of the continuous sinograms that have identical phases of the motion effects. The CSS method provides phase coherent sinograms bases on the SOC tracking of organ motion, or
- Third, a retrospective gating scheme that is based on the waveform output results of the SOC that tracks the cardiac motion. This scheme identifies the phases of motion, when the heart is moving the least, that are used to synthesize a 180 degrees plus the fan angle sinogram for image reconstruction.

Practical applications of the novel signal processing concepts of this paper in medical diagnostic imaging applications have been identified to include detection of cardiac blood vessel calcification and imaging of the ventricles. Thus, the reported real results in this paper have been focused on cases that demonstrate the effectiveness of the above method as a simple screening process to detect cardiac blood vessel calcification. It is anticipated that this approach would lead to the development of a CT scanner dedicated to assist the above screening process, which cannot be accomplished by the current ultrasound or MRI medical tomography imaging modalities.

II. THEORETICAL REMARKS

II.1 Definition of Parameters

The data acquisition process in CT applications is depicted in Figure 1. The projection measurements, $\{p_n(r_n, \theta_i), (n = 1, \dots, N)\}$, shown schematically in this figure, are defined as the line integrals along lines passing through the image $f(x, y)$. For a given detector n , and projection angle θ_i , $p_n(r_n, \theta_i)$ is given by:

$$p_n(r_n, \theta_i) = \iint f(x, y) \delta(x \cos \theta_i + y \sin \theta_i - r_n) dx dy, \quad (1)$$

where $f(x, y)$ is the tomographic cross section that we wish to reconstruct. Equation (1) is applicable for parallel beam projection CT scanners. The angular step increment between two successive projections of the X-ray scanner is defined by: $\Delta\theta = 2\pi/M$, where M is the number of projections taken during the period T required for one full rotation of the source and receiving array around the object $f(x, y)$ that is being imaged. For fan beam scanners, the projection function includes the following transformations,

$$r_n = R \sin \sigma_n, \quad \theta_i = \sigma_n + \beta_i, \quad (2)$$

to account for the geometry of the fan beam, as shown in Figure 1. The projection function is now defined by [13]:

$$g_n(\sigma_n, \beta_i) = p_n([r_n = R \sin \sigma_n], [\theta_i = \sigma_n + \beta_i]). \quad (3)$$

In the image reconstruction process the pixel $I(j, k)$ in the actual image, shown at the right of Figure 1, corresponds to the Cartesian point (x_j, y_k) in the CT scan plane. The pixel value $I(j, k)$ is given by:

$$I(j, k) = \sum_{n=1}^M G_n(\sigma_n, \beta_i), \quad (4)$$

where $G_n(\sigma_n, \beta_i)$ is the filtered version of the projection $g_n(\sigma_n, \beta_i)$ that has been adjusted to account for geometric effects. The filtering function used in this investigation is the Ram-Lak filter [11], cascaded with the Parzen window. The angle σ_n , which defines the detector that samples the projection through a

point (x_j, y_k) , for a given projection angle β_i , is provided by Equation (5), where (r_d, φ) is the polar representation of (x_j, y_k) .

$$\begin{aligned}\sigma_s &= \tan^{-1} \left[\frac{r_d \sin(\beta_i - \varphi)}{R + r_d \cos(\beta_i - \varphi)} \right], & \frac{-\pi}{2} \leq (\beta_i - \varphi) < \frac{\pi}{2} \\ \sigma_s &= \tan^{-1} \left[\frac{r_d \sin(\varphi - \beta_i)}{R + r_d \cos(\varphi - \beta_i)} \right], & \frac{\pi}{2} \leq (\beta_i - \varphi) < \frac{3\pi}{2}.\end{aligned}\quad (5)$$

II.2 The Spatial Overlap Correlator Concept

In SONAR applications [8,9], it has been shown that the spatial overlap correlator increases the angular resolution (array gain) and reduces artifacts that are caused by scattering and, the target's or receiving-array's motion effects. The approach of generating a synthetic aperture is based on computing an appropriate phase correction factor to coherently synthesize spatial measurements [8] into an equivalent longer physical aperture size. Extending this scheme to the case of a CT application is not straightforward. The computation of the correction factor is still based on the comparison of spatially overlapping measurements at two consecutive time moments of the CT sampling process. This factor compensates for the phase or amplitude fluctuations caused by the subject's organ-motion effects and tracks the information due to the movement of organs such as the heart and lungs.

Shown in Figure 2, is the concept of the spatial overlap correlator for CT applications. In a sense, this is a data acquisition scheme to obtain an estimate of the motion information for subsequent removal. As shown in Figure 2, the scheme uses two sources, and an array receiver. The system of two sources - array receiver performs the data acquisition process in a conventional manner. Specifically, at time t_0 , which is the starting time of the CT data acquisition process, the array receiver and the two sources are at spatial locations s_1 and s_2 respectively, as shown by the dark dots in Figure 2. At time $(t_0 + \Delta t)$, where $\Delta t = T/M$, is the elapsed time between two successive firings of the X-ray, the two sources will be at locations s_2 and

s_3 , shown by the lighter dots in Figure 2. The process is repeated throughout the data acquisition period T of one full rotation.

Based on the schematic representation of Figure 2, the CT measurements corresponding to the spatial location s_2 are obtained twice at time t_0 and $(t_0 + \Delta t)$. This is equivalent as taking two consecutive photographs at time t_0 and $(t_0 + \Delta t)$ of a field of view that includes a moving train and other stationary objects, such as trees and houses. If the images of these two photographs are subtracted at the pixel level, then the resulting image will include information only about the moving train. Based on this simplistic interpretation of the information provided by the two sources CT measurement system, as depicted in Figure 2, a comparison (or correlation) of the two spatially overlapping measurements will provide a new set of measurements that should describe any type of organ motion that may have occurred during the time increment Δt .

More specifically, suppose

$$\{ p_{n_1}(r(t), \theta(t), t_0), (n_1 = q, q+1, \dots, N), \} \text{ and } \{ p_{n_2}(r(t), \theta(t), t_0 + \Delta t), (n_2 = 1, 2, \dots, N-q), \} \quad (6)$$

are the $(N-q)$ spatially overlapping measurements for the projection $\theta(t)$ received by the N -element array at the two successive moments t_0 and $(t_0 + \Delta t)$, as depicted in Figure 2. A time dependency has now been included in all the parameters of Equations (1) and (2). Based on Equation (2), the time dependent projection measurements for a fan-beam scanner is expressed by:

$$g_n(\sigma(n), \beta(t), t_0) = \iint f(x, y, t) \delta\{x \cos[\sigma(n) + \beta(t)] + y \sin[\sigma(n) + \beta(t)] - R \sin \sigma(n)\} dx dy, \quad (7)$$

where $f(x, y, t)$, is a time varying tomography object. For CT applications, the amplitude difference between the two sets of measurements of equation (6) is defined by,

$$\Delta p_{n_0}(r(t), \theta(t), t_0 + \Delta t) = p_{n_2}(r(t), \theta(t), t_0 + \Delta t) - p_{n_1}(r(t), \theta(t), t_0), \quad \text{for } (n_0 = 1, 2, \dots, N-q). \quad (8)$$

For fan-beam scanners, the difference $\Delta g_{n_0}(\sigma(n), \beta(t), t_0 + \Delta t)$, is defined by:

$$\Delta g_{n_o}(\sigma(n_o), \beta(t), t_o + \Delta t) = g_{n_{s2}}(\sigma(n_{s2}), \beta(t), t_o + \Delta t) - g_{n_{s1}}(\sigma(n_{s1}), \beta(t), t_o) \quad \text{for } (n_o = 1, 2, \dots, N-q) \quad (9)$$

where, $(n_{s1} = q, q+1, \dots, N)$, and $(n_{s2} = 1, 2, \dots, N-q)$. As a result of equation (7), and recalling the spatial overlap concept of equation (6), which assumes that $\{\beta_{s1}(t_o + \Delta t) = \beta_{s2}(t_o)\}$, $\{\sigma(n_{s1}) = \sigma(n_{s2})\}$, Equation (9) takes the form:

$$\Delta g_n(\sigma(n), \beta(t), t_o + \Delta t) = \iint [f(x, y, t_o) - f(x, y, t_o + \Delta t)] \{ \delta \{ x \cos[\sigma(n_{s2}) + \beta_{s2}(t_o)] + y \sin[\sigma(n_{s2}) + \beta_{s2}(t_o)] - R \sin \sigma(n_{s2}) \} dx dy, \quad (10)$$

where $[f(x, y, t_o) - f(x, y, t_o + \Delta t)]$ indicates the tomographic differences of the time varying object between the time moments t_o and $(t_o + \Delta t)$. From this arrangement, for a stationary image, it is evident that this two source configuration will provide spatially identical sets of measurements at all spatial locations, despite being taken at two successive moments in time, separated by Δt . If the measurements of the spatial overlap correlator depicted in Figure 2 and defined by Equations (9) and (10) are provided at the input of an image reconstruction algorithm, then the resultant image will show the time dependent changes $[f(x, y, t) - f(x, y, t + \Delta t)]$ (for $t = t_o, \dots, t_o + T$) that took place during the time period T of the CT data acquisition process. This suggests that the image reconstruction process of the spatially overlap correlator measurements of Equation (10) will track only the moving components of the object. The stationary components will have zero contribution and they will not appear in the reconstructed image. This is confirmed by the results of Figure 4, which are discussed later. Figure 3, presents a graphical representation of the above two dimensional (2-D) space-time sampling process defined by Equations (9) & (10) and Figure 3. The vertical axis shows the time moments associated with the angular positions of the source-array receiver that are shown by the horizontal axis. The sets of line-segments along the diagonal represent the measurements of a CT scanner at different angular positions. Image reconstruction algorithms assume

stationarity in the image, or that these sets of measurements are along a horizontal line of Figure 3 (i.e. that they have been acquired at the same time moment).

However, as shown in Figure 3, during the period of T seconds of the scanner's one full rotation of data acquisition period, the image of interest may suffer deformation as is the case of organ motions (i.e. heart, lungs, etc.). In the same figure, the spatially overlapping measurements of Equations (9) and (10), are depicted by the pair of lines that overlap in angular space but are taken at two successive time moments. This process is equivalent to a synthetic aperture processing scheme for SONAR applications [8,9]. An obvious organ motion correction process is to remove the temporal amplitude and phase differences, due to organ motion effects, between the spatially overlapping sets of measurements. Ideally, this correction process will correct the measurements with respect to the first measurement, which is equivalent in moving the diagonal measurements into the same instant in time and along the horizontal line, shown in Figure 3, that corresponds to the time moment t_0 . This approach has been tested with synthetic data and it has been shown that it is not effective because the number of projections, M , in CT scanners is very large $O(10^3)$. As a result, the propagation of errors that are generated by the correction process build up significantly and destroy useful information.

At this point it is important to emphasize the phase alignment between the measurements provided by the spatial overlap correlator (SOC) that tracks the motion effects with respect to the phase corresponding to the conventional CT data acquisition process. All these issues and a physical interpretation of the measurements provided by the SOC and the conventional CT, are discussed in detail in [12,13]. Briefly, the results in [12,13] show that the measurements provided by the SOC are the derivative of the difference of the sinograms corresponding to the tomography image with the organ motion effects and those acquired by a CT system without the motion effects. It is shown [12,13] also that the measurements of the spatial overlap correlator form the basis of an optimum adaptive processing scheme to remove motion artifacts associated with the CT data acquisition process.

II.3 Simulation Results for the Functionality of the Spatial Overlap Correlator

The functionality of the spatial overlap correlator was simulated with the Shepp-Logan phantom as a test image [11]. The software consisted of four signal processing components:

1. the projection data generation based on a test phantom,
2. the image reconstruction algorithm, which was the filtered back-projection method including a Ram-Lak filter cascaded with the Parzen window,
3. the SOC signal processing concept described in this paper, and
4. the image display.

The geometry in the simulations and the rest of the parameters were similar with those of Siemens x-ray CT Somatom 4 medical imaging system. In particular, the simulations included $T = 0.75$ seconds period of rotation for the CT system with a set of $M=1056$ projections by a $N=768$ array of detectors. This is because the spatial overlap correlator has been tested with real data sets from the above real time system provided by Siemens.

The left image in Figure 4 shows the reconstructed Shepp-Logan phantom for stationary case. No motion artifacts are present in this image. The center image in Figure 4 includes motion at the left ellipse, which is taking place during the data acquisition period of the CT scanner. The rest of the components of the phantom were assumed stationary. Thus, motion artifacts are evident in this case only at the left ellipse of the reconstructed phantom. Finally, the right image in Figure 4 shows the image reconstruction output of the spatial overlap correlator. As expected from Equation (10), the reconstructed image from the SOC output includes the moving ellipse only of the Shepp-Logan phantom. None of the stationary components of the phantom appear in this image, which may be of important diagnostic value to track organ motion effects, and further may be used in processing schemes for motion artifact removal [12,13].

III. SYSTEM IMPLEMENTATION

An essential technical requirement by the spatial overlap correlator is the implementation of two X-ray sources separated with an angular spacing equal to an integer number of the sensor spacing of the receiving array, as shown in Figure 3. However, because of the massive size of an X-ray source, the two-source concept for a CT scanner is technically not feasible. In this investigation, it is suggested that the second source can be generated by the deviation of the x-ray beam, which forms the flying focal spot functionality of Siemens Somatom-4, Elscint's Twin RTS CT scanners and GE's wobbling focal spot. The new active transmission of the wobbling or flying focal spot functionality defines a new source at an angle $\delta\theta$ and time instant $\Delta t/2$ with respect to the previous source position. Typical values for $\Delta t/2$ are approximately 0.5ms. This process provides a second set of spatial samples that are offset by $\frac{1}{2}$ sensor spacing in the receiving array, hence it is equivalent to doubling the spatial sampling frequency of a CT fan-beam scanner.

Thus, an implementation effort of the spatial overlap correlator concept in CT systems may consider the flying focal point functionality that can be modified in such a way that will simulate the second source being activated about 0.5ms later than the previous X-ray transmission. However, the source should be deviated in such a way that the second active transmission should be formed approximately at the same location as the previous one and with an appropriate re-alignment of the detector array. It has been assessed that this kind of implementation effort requires costly hardware modifications in existing CT scanners that include the flying focal spot functionality. During the course of this investigation, data from a CT system including the hardware implementation of the spatial overlap correlator have not been available to assess the effectiveness of the AIC process. Therefore other practical alternatives had to be investigated to address this kind of implementation issues.

III.1 Generic Software Implementation of Spatial Overlap Correlator in CT Systems

Based on the concept of the Spatial Overlap Correlator in Figure 2, a generic implementation of the SOC is now defined. This generic approach is a software implementation that requires no hardware modification of the CT scanner. As a result, this software implementation is described as the Software Spatial Overlap Correlator (SSOC).

In this approach the fact that the sampling process by the x-ray CT scanner is circular is exploited. In other words, samples at time t and $t+T$ are of identical spatial locations. Therefore, samples separated by a time interval T are compared in the software implementation of the spatial overlap correlator.

For simplicity, but without the loss of generality, let us consider the case of a sinusoidal variation. As described by Equation (11), the difference between two sinusoidal varying signals, of the same frequency, but different phase yields a new sinusoidal signal also of the same frequency, but different phase. This is provided that the phases of the two signals are not equal, and that their difference in phase does not correspond to an integral number of periods.

$$\sin(2\pi f_0 t) - \sin(2\pi f_0 (t + t_0)) = -2 \left[\sin(\pi f_0 t_0) \cos \left(2\pi f_0 \left(t + \frac{t_0}{2} \right) \right) \right] \quad (11)$$

It is evident from Equation (11) that the difference of a periodic signal and a delayed version of this signal yields a signal of the same frequency and different phase, as expected. This relationship holds if the product $(f_0 t_0)$ is not an integer. This means that the signal is not delayed an integral number of cycles. When the product $(f_0 t_0)$ is an integer, corresponds to the case of the periodicity of the motion corresponding exactly to the periodicity of the sampling of the motion. In this case, the SSOC does not track the motion, and the system appears stationary. In cardiac x-ray CT imaging applications, this occurs in cases where the period of the subject's cardiac cycle coincides exactly with the period of data acquisition of the x-ray CT scanner.

IV. MOTION CORRECTION SCHEMES BASED ON WAVEFORMS OF SSOC

Equation (11) suggests that the software spatial overlap correlator is an effective technique to track the phases of the organ motion. More specifically, analysis of the time series of a single detector from the spatial overlap correlator provides information about the phase of the motion of the moving object or organ. In other words, the SSOC waveforms (or time series) give the position of the object at the time that each view is taken. Thus, the SSOC waveforms are equivalent with those of an ECG system, which under appropriate synchronization with the acquisition process of x-ray CT scanners provide retrospective or prospective ECG gating to reduce motion artifacts in cardiac imaging applications. This kind of equivalence between the SSOC and ECG waveforms forms the basis for the two motion corrections schemes of this investigation, the retrospective gating (RG-SSOC) and the coherent sinogram synthesis (CSS) schemes.

IV.1 Retrospective Gating Based on Waveforms of Spatial Overlap Correlator (RG-SSOC)

Morehouse et al. [14] introduced retrospective gating in 1980, as a motion correction technique for x-ray CT cardiac imaging applications. The technique included acquisition of projection data continuously during a few rotations of the x-ray source and selecting the projection data acquired during a specified window of the cardiac cycle, during diastole when the heart is moving the least. Thus, system integration of an ECG system with the x-ray CT scanner was an essential requirement in this technique to monitor the cardiac motion. In 1983, Moore et al. [15] used prospective ECG gating to determine the optimal start-time for each rotation, later including coincident-ray considerations [16]

Since the waveforms of the spatial overlap correlator scheme track the cardiac motion, as shown later in Figure 13, they can be used to identify the phases of motion, when the heart is moving the least, in order to synthesize a 180 degrees plus the fan angle sinogram for image reconstruction. This leads to a

retrospective gating (RG) scheme that is based on the output results of the spatial overlap correlator (RG-SSOC). In this investigation, the RG-SSOC approach of gating has been assessed and compared with the established ECG-gating process and the relevant results from clinical testing with human patients are summarized in Section V.2.1.1. Thus, the new gating approach may eliminate the requirement for system integration of ECG systems with x-ray CT scanners for cardiac imaging applications.

IV.2 Coherent Sinogram Synthesis from the Software Spatial Overlap Correlator

Analysis of the time series of a single detector from the spatial overlap correlator provides information about the phase of the motion of the moving object or organ. In other words, the time series give the position of the object at the time that each view is taken. This information is the basis for the coherent sinogram synthesis (CSS) scheme. The measurements from the overlap correlator make it possible to identify the phases of the motion at each time moment, thus allowing the construction of a new coherent sinogram that is synthesized using only the views that correspond to a single point in the phase of the motion. Image reconstruction on this synthesized data set, or synthesized sinogram, produces an image that depicts the moving object frozen at a single point in its phase. Furthermore the CSS process does not require any kind of periodicity for the motion effects. It requires, however, repetition of the selected phase of motion.

The next step is to assemble a sinogram using the coherent sinogram synthesis method, which uses the time series produced by the spatial overlap correlator. The technique requires first selection of a specific phase of the detected motion cycle by the SOC process and defines this as the phase of interest. It then isolates every subsequent time moment during the data acquisition period when the object is again at the pre-selected phase of its motion cycle. A number of projections is selected at each of these time moments, and assembled into a phase coherent sinogram.

For the purpose of illustration, Figure 5 depicts the process using the curves obtained from the output of the software spatial overlap correlator process. When the periodicity of the organ motion is established from a set of two CT rotations and through the measurements of the SSOC, then an internal triggering will select the corresponding CT measurements with the required in-phase information of the organ motion to synthesize coherently a new sinogram, thus minimizing a large number of continuous rotations and the X-ray radiation level.

IV.2.1 Phase Selection by Correlation

Selection of the in phase information of the detected motion cycle by the SOC process is provided by a sliding window correlation processing, which isolates the moments during the scanner's data acquisition process. In general, the cross-correlation coefficient CC_{sr} between two time series, s_i and r_i of length L is given by Equation (12).

$$CC_{sr} = \frac{\sum_{i=1}^L s_i r_i}{\sqrt{\sum_{i=1}^L s_i^2 \sum_{i=1}^L r_i^2}} \quad (12)$$

The sliding window correlation technique uses a subset of the signal near the phase of the interest r_i as the replica, and then correlates this replica with segments of the continuous signal s_i to compute a time varying correlation function. The segments of the signal used in the cross-correlation function are selected in a sliding window fashion. The time varying cross-correlation function CC_i is given by Equation (13), where L is the length of the segment used in the cross correlation function, N is the length of the complete time series from the spatial overlap correlator.

$$CC_i = \frac{\sum_{j=-L/2}^{L/2} s_{i+j} r_{j+L/2}}{\sqrt{\sum_{j=-L/2}^{L/2} s_{i+j}^2 \sum_{j=-L/2}^{L/2} r_{j+L/2}^2}}, i = \frac{L}{2}, \frac{L}{2} + 1, \dots, N - \frac{L}{2} \quad (13)$$

The time varying correlation function for a selected phase of interest is shown in the lower curve of Figure 5. The upper curve shows the smoothed time series from the software spatial overlap correlator. The phase of interest is marked on this curve. Using a replica that is centered at the point marked as the phase of interest, the time varying correlation function is shown in the lower curve. From the time varying correlation function, it is evident that the level of correlation between the replica and the upper curve, which describes the organ motion, reaches a maximum value of approximately one whenever the phase of interest reoccurs. The vertical bars in the lower curve of Figure 5 indicate the detected maxima. The time moments that the value of these maxima are in the range of $(0.8 < CC_i < 1.0)$ then are considered as the time moments at which the phase of interest reoccurs. The time moments directly define a projection number, since the projections are acquired sequentially at a known sampling rate. Then, a coherent sinogram may now be assembled by using all the above projections that define the selected phase of organ motion of interest.

IV.2.2 Assembling Coherent Sinograms

The curve in the lower segment of Figure 6, which is similar to the lower panel of Figure 5, shows the time moments selected because of the sufficiently high level of correlation. These time moments map to angular locations determined by the physical location of the x-ray CT source and detector array at the time of the selected phase of the periodic organ motion, as shown in the upper panel of Figure 6. Since the data acquisition process of an x-ray CT scanner is circular, all of the spatial locations, shown in the upper part of Figure 6, are sampled repeatedly. A projection number P maps to a physical projection number p , where P is referenced to the entire data acquisition period, and p is referenced to the physical location on the circle around which data is acquired by the x-ray CT system. The relationship between P and p is given by the circular mapping in Equation (14), where M is the number of projections taken by the x-ray CT scanner in a single rotation.

$$P = mM + p; 0 \leq P \leq \infty; 0 \leq p \leq M; m = 1, 2, \dots, \infty \quad (14)$$

In effect the selected segments of a sinogram are synchronized to the organ motion cycle, in that valid projections are only obtained when the organ is exactly at the desired point in its motion cycle. Under ideal conditions, one view will be taken at each time moment that the organ reaches the desired point in its motion cycle; and after N cycles of motion a complete sinogram will be available. Since there is no physical synchronization between the data acquisition process and the organ motion, there may be repetition of some views, while some views may still be missing.

Figure 6 and Equation (14) underline this fact that it may be impossible to generate a complete sinogram using the CSS method. Thus, an interpolation process is required to account for any missing angular segment, as those shown in the upper panel of Figure 6.

IV.2.3 Interpolation

Since the CSS method may be considered as a data acquisition process that is both temporally and spatially varying, the use of standard interpolation methods is not an option. In general, interpolation methods such as the cubic-spline interpolation [17], will be successful at interpolating large structures which are well defined in the sinograms, but will not interpolate small spatial structures, especially those associated with cardiac imaging applications.

The interpolation techniques used in the coherent sinogram synthesis (CSS) method take data from one complete revolution of the CT system and use it as a basis for the coherent synthesis of the sinogram. The idea is to try to improve the image that would be produced by this standard x-ray CT sinogram. Thus, using the conventional sinogram as a starting point, fixed sized projection windows are selected in the same manner as described in the previous section. Whenever a suitable projection window is found, it is used to overwrite the original projections of the conventional sinogram. This method is the base image

approach since it starts with a standard x-ray CT image, and it is useful if the initial image quality is sufficiently good, since it modifies the original sinogram in an attempt to obtain an improved image.

Another approach is to remove any limit on the number of projections in the selected sinogram segments, and may be considered as a flexible window approach. Thus, the size of the selected segments of projections are allowed to be as large as necessary to fill in the entire sinogram. The center of the initial segment of projections is defined as an ideal point. The missing projections are then filled in using appropriate projections from the data acquired so that each projection is as close as possible to an ideal point. This method is preferred when the original image quality is poor. Once a coherently synthesized sinogram for a single phase of the organ motion is complete, the image is obtained by conventional image reconstruction processing, which is equivalent to that of a stationary object.

IV.2.4 Motion Picture

A direct benefit of the CSS method is the ability to generate phase coherent motion pictures. A number of images are generated, each at a different point in the phase of the motion of the object. A number of phases of interest are selected, equally spaced over a single cycle of the motion cycle; and an image is created for each of these points. A typical example of reconstructed phases of motion are shown later in Figure 11. These images in the proper order will produce the frames of an X-ray CT motion picture that shows the moving object as it goes through the various phases of its motion.

However, the coherent sinogram synthesis (CSS) method requires that a data set should be acquired over a number of revolutions of the X-ray CT scanner, as opposed to just one revolution in straightforward x-ray CT image reconstruction. The number of revolutions required is a design parameter of the algorithm. For the CSS method, this number is simply a function of desired image quality. There is a direct relationship between the length of the data sequence acquired and the correctness of the synthesized sinogram to the equivalent sinogram of a stationary object. This correctness is also a function of the speed of the object's

motion. Generally, a faster moving object will yield a more exact sinogram than a slow moving object for a fixed length data sequence, because there are more cycles available. Results from the RG-SSOC and CSS-SSOC methods with real data sets are described in the next section.

V. EXPERIMENTAL RESULTS

Two sets of experiments were performed using a Siemens Somatom 4 Plus X-ray CT scanner. The initial experiments were performed using a moving phantom. After validation of the method with these results, further tests were carried out with human subjects. The target application of the retrospective gating and the coherent sinogram synthesis methods are in the area of cardiac x-ray CT imaging. Therefore the experiments with human subjects attempted to apply the techniques to data sets that were acquired by taking cardiac cross-sections from a number of human patients.

V.1. Phantom Experiments

The first sequence of experiments was carried out using a moving phantom in the Siemens Somatom Plus 4 x-ray CT scanner. Figure 7 shows the phantom configuration that was constructed with a hollow Plexiglas cylinder and a solid Teflon cone inside. The cone and cylinder share a common axis. Also seven metal wires were placed on the outside of the cone, some straight and along the axis of the cone, and other in spirals along the surface of the cone. The wires, which are between 0.5 and 1.0 mm in diameter, are not easily visible in Figure 7. In the view of the scan plane of the x-ray CT system, the reconstructed image of the stationary phantom appears as a circular shell with a shaded circle inside, and a number of dots that are due to the wire cross-sections. This is shown in Figure 8. The light circle in the center of the phantom is the metal axis along which the structure was moved, and the foam mattress is visible as a rectangle below the phantom. The image of Figure 8, is the output of the CT scanner for the stationary phantom and represents the ideal image. Since there is no motion present, there are no motion artifacts. Thus, this

image may be considered as the target image to assess the performance of the SSOC and CSS processes from this experiment.

To simulate motion, the cylinder and cone combination was moved back and forth along the axis of the structure by an electrical motor. Moving the phantom back and forth simulates the deformation of the inner circle, and translational motion of the metal wires. The speed of the motor defines the period of the inner circle's deformation. The position of the attachment to the flywheel of the motor defines the displacement of the phantom, and consequently the deformation amplitude of the motion in the scan plane. A number of experiments were carried with varying frequencies of motion. The period of motion was approximately 0.6 second (1.67Hz), 0.75 second (1.33 Hz), 1.2 seconds (0.83 Hz) and 2.0 seconds (0.5 Hz). The CSS technique based on motion tracking by software spatial overlap correlator was implemented in the same manner as described in Section IV.2. The processing and results of these experiments carried out are discussed in detail in the following section.

V.1.1 Experiment: Moving Phantom, 0.6 Second Period of Motion

The first experiment was carried out with motion period of 0.6 second. This corresponds to 1.67 Hz, or 100 beats per minute in terms of cardiac motion. The image obtained using standard x-ray CT techniques is shown in Figure 9. The degradation in the quality of the image, compared to the image under stationary conditions in Figure 8, is obvious. The outer circle that is due to the cylinder is undistorted. This is because the motion of the cylinder is along its axis, hence it appears stationary in the scan plane. The same is the case with the foam mattress, which is not moving. However, the filled inner circle and the dots that are due to the wires are completely distorted. There is no evidence that the inner structure should be circular, or that there should be stationary dots where the wires are positioned.

The motion time series extracted by the software spatial overlap correlator are shown in the upper curve of Figure 10. The same time series after smoothing (low pass filtering) is shown in the second curve from the

top. The properties of the motion of the phantom are immediately obvious in these curves. In fact it is possible to accurately compute the actual period of the motion from the motion curves, which is 0.63 second. The time varying correlation function for one selected phase of interest, which is selected to be the peaks of the motion time series, is shown in the third curve from the top. The vertical bars in this curve indicate the local maxima of the time varying function. The vertical bars in the bottom curve show the projections that are selected as the center of segments for coherent sinogram synthesis. The flexible window interpolation scheme was used in this experiment, as well as in the others with this phantom.

Figure 11 shows six images generated at six equally spaced points in the phase of the motion cycle, according to the process discussed in Section IV.2. These six images show one complete motion cycle. The images that correspond to points in the phase where the object is at stationary points of its motion cycle show the details more clearly. Comparison of the top-left image of Figure 11 and the ideal image in Figure 8 yield no major differences. The boundaries of the objects, especially the cone, and the position of the metal wires are clear and well defined. The boundaries of objects, and the positions of the metal wires in intermediate phases, where the velocity of the object is high, are not as clear. Despite the fact that some phases are not as clear, all six images from the coherent sinogram synthesis method are superior to the image in Figure 9 that was obtained using standard x-ray CT techniques. The corrected images describe the object more accurately than the standard x-ray CT image.

Since the RG-SSOC process is applicable to type of motions that include a phase with the least movement, the RG-SSOC process cannot be applied on the projection measurements associated with the phantom, because of the continuous movement of the cone in the phantom. This indicates the primary difference in terms of applications between the RG-SSOC and the CSS-SSOC methods. In other words, the CSS-SSOC method is best suited for a type of motion that is continuous and reproducible.

V.2 Results with Human Patients

Performance assessment of the CSS-SSOC and RG-SSOC processes with human patients was carried out at the Oncology Clinic of the Hospital of Offenbach-Germany with a Siemens Somatom Plus 4 x-ray CT scanner. The CT/X-ray data sets from human patients, that were used in our performance assessment study, were part of a brachytherapy procedure for cancer treatment that included high doses of radiation. Brachytherapy procedures require that the tumor boundaries should be established with CT scanners and if possible with 3D ultrasound systems. Thus, the clinical procedures of this investigation to assess the effectiveness of the CSS-SSOC and RG-SSOC processes for detecting/screening cardiac blood vessel calcification were preliminary. Currently, the Oncology Clinic of the Hospital of Offenbach-Germany and the Department of Diagnostic Radiology of the Klinikum Grosshadern University of Munich-Germany, members of our New Roentgen project consortium [20], are in a process to establish clinical procedures for detecting/screening healthy people for cardiac problems, using CT scanners without ECG triggering. The starting point to establish these new clinical procedures could be the RG-SSOC process discussed in Section IV.2 and demonstrated with real results in the following Section V.2.2. The results corresponding to the CSS-SSOC and RG-SSOC with human patients, carried out under this investigation, are discussed in detail in the following sections.

V.2.1 Human Patient #1: 75 bpm cardiac motion

V.2.1.1 Motion Correction Using the RG-SSOC and Retrospective ECG Gating Methods

The objective in this set of clinical testing was to assess and compare the RG-SSOC method with respect to the established retrospective ECG-gating process. Thus, the data acquisition process in this case required synchronization between the ECG waveforms and the starting point of recording of projection data from the x-ray CT scanner. The CT period of rotation, which was set to 1.0 second. Figure 12, shows the signal waveforms that allowed this kind of synchronization process. The upper part of Figure

12 presents an ECG waveform and the lower part the triggering signals as square pulses that were sent from the ECG system to the x-ray CT scanner. The first square pulse signaled the start of the x-ray CT acquisition process of projection data. Figure 13 shows the synchronized outputs of the SSOC and ECG triggering signal that allowed the comparison of both gating methods with the same set of data from the human patient #1. During the CT data acquisition process the patient was asked to follow the breath holding procedures. The CT was positioned to obtain the ventricles of the heart's lower part.

The upper part of Figure 13 shows the output of the SSOC that indicates a complex spectrum of frequencies. The middle part of Figure 13, shows the band-pass filtered output of the SSOC waveform centered on the frequency of the human patient's heart beat rate of 75 bpm.

The lower diagram of Figure 13 shows the ECG triggering signals that were used as a reference to choose the segments of the sinogram for retrospective ECG gating. These segments of the sinograms were chosen from a section of the ECG waveform located between two consecutive QRS-peaks (and the corresponding two consecutive square waves of the ECG triggering signal) and to coincide with the range between the T and P phases of the ECG-waveform, as shown in Figure 12. The section between the T and P phases of the ECG signal represents the diastole phase, when the heart is moving the least. This section is then used to synthesize a 180 degrees plus the fan angle sinogram for image reconstruction. The vertical lines labeled as Proj#---, crossing all the three waveforms, indicate the starting point of the sinogram-segment with temporal length corresponding to 180 degrees plus the fan angle.

The properties of the patient's heart motion are immediately obvious in these curves. In fact it is possible to compute the actual period of the motion from the motion curves, which is approximately 0.73 second. Moreover, the SSOC time series reveal that the amplitude of the patient's heart movement increases significantly during the CT data acquisition process. This is an indication that the patient was not able to maintain the breath-hold and started to gasp.

It is evident from the middle SSOC waveform that there are two very close frequencies in the spectrum separated by approximately 0.19-Hz, as indicated by the beating effect. Since the SSOC waveform in this case tracks the motion of the heart's two ventricles, it is evident by the presence of the two close frequencies that the two ventricles of the human patient #1 are out of sync. This is an observation that is clinically correct, since the human patient's weight was above 150-Kgr and his age was 65 years old.

Figure 14 presents the output of conventional CT scanner for the human patient #1, with breath holding. The tomography area of this image shows the lower part of the heart's ventricles. The results of motion correction using the retrospective ECG-gating method are shown in Figure 15. The projection point #6045, shown in Figure 13, defines the range between the T and P phases of the ECG-waveform and the starting point of the sinogram-segment with temporal length of 180 degrees plus the fan angle. The same projection point #6045 defines the area of the SSOC waveform (upper curve in Figure 13) with the least cardiac motion. Minimization of motion effects in this case allows identification of the heart's pericardium and its thickness, at the upper part of the image. The circular lobe at the left part of the image shows the tomography of the top part of the liver.

To demonstrate the effectiveness of the SSOC waveform and its accuracy with respect to the ECG waveforms to identify the areas with minimum or maximum motion effects in the sinogram, Figure 16 presents the image reconstruction from a sinogram-segment gated at the projection point #26202, shown in Figure 13. This starting point of the sinogram-segment with temporal length of 180 degrees plus the fan angle defines the area of the SSOC waveform (upper curve in Figure 13) with large amplitude motion effects due to patient's gasping. This observation suggests that the motion artifacts should be present in the corresponding reconstructed image. However, the ECG signals (lower curve in Figure 13) suggest that the same projection point #26202 should provide a retrospective ECG-gating corrected image with minimum motion effects. However, these motion effects, which are evident in the reconstructed image in Figure 16 with inferior image clarity than that in the previous image in Figure 15, suggest that the SSOC

process is very accurate to identify characteristics of the heart's localized motion (close to the location of the CT slice) that cannot be detected by an ECG waveform.

Figure 17 shows the image results of motion correction using the RG-SSOC gating at the projection point #4323, shown in Figure 13. According to the ECG signals presented at the lower part of Figure 13, this projection point (eg. #4323) includes the QRS phase of the ECG-waveform, which suggests the strongest heart motion effects. For the same projection point #4323, however, the SSOC waveform (upper curve in Figure 13) indicates very small amplitude motion for the localized section of the lower part of the heart's ventricles. Motion effects in this corrected image (Figure 17) are not so prominent as in Figure 16. In fact the thickness of the heart's pericardium as well as the boundaries of the ventricles can be identified with the same or better clarity as in Figure 15.

To verify the consistency of the SSOC process in identifying the sections of the SSOC waveforms with the minimum motion effects, the image in Figure 18 shows the motion correction using the retrospective RG-SSOC and ECG gating methods from a set of projection data from the human patient #1 with free breathing and with contrasting agent. The starting projection point for gating was chosen from the characteristics of the SSOC and ECG waveforms, both showing very small motion amplitude, as in the case of projection#6045, in Figure 13. Motion effects in this case are not so prominent as in Figure 16. The thickness of the heart's interior walls (pericardium) as well as the boundaries of the ventricles can be identified with the same or better clarity as in Figures 15 and 17.

V.2.1.2 Motion Correction Using the CSS Method

The coherent sinogram synthesis (CSS) process, as defined in Section IV.2, was applied on the output results of the SSOC process, shown in Figure 13. The corresponding reconstructed image for human patient#1, using the CSS-SSOC method, is shown in Figure 19. The corrected image in this case suggests that the CSS-SSOC method has not been successful, as in the case with the moving phantom, to minimize the motion artifacts to the same degree as with those of Figures 15, 17 and 18 using the RG-SSOC and

ECG gating processes. An explanation for this failure regarding the performance of the CSS-SSOC method is that it requires well defined continuous motion characteristics, which are reproducible and detectable by the SSOC process. The same kind of performance characteristics for the CSS-SSOC method will be identified also with the next set of data with human patients.

V.2.2 Human Patient #2: 95 bpm cardiac motion

The data acquisition arrangements for the CT data sets from the human patient#2 were similar to those of the moving phantom experiments and the human patient#1, except for the CT period of rotation, which was set to 0.75 seconds. In this case, no ECG signal was recorded. The patient's pulse rate was approximately 95 beats per minute. During the CT data acquisition process the patient was asked to follow the breath holding procedures.

The motion time series extracted by the software spatial overlap correlator are shown in the upper curve of Figure 20. The same time series after smoothing (low pass filtering) is shown in the second curve from the top. The properties of the heart motion of the patient are immediately obvious in these curves. In fact it is possible to compute the actual period of the motion from the motion curves, which is approximately 0.65 second. Moreover, the SSOC time series reveal that the amplitude of the patient's heart movement increases significantly during the CT data acquisition process. This is an indication that the patient was not able to maintain the breath-hold during the 20-seconds period and started to gasp, which is the same situation with human patient#1.

The time varying correlation function for one selected phase of interest is shown in the third curve from the top of Figure 20. The vertical bars in this curve indicate the local maxima of the time varying function. The vertical bars in the bottom curve show the projections that are selected as the center of segments for coherent sinogram synthesis. The base image interpolation scheme was used to coherently synthesize a sinogram for the selected phase of the heart movement. The top image in Figure 21 shows the

conventional x-ray CT image for subject #2. The tomography image of the heart is at the centre. The lung structure characteristics are also shown at both sides of the heart image. Moreover, the same CT image shows a large tumor at the left side of the heart.

There are no strong indications of heart motion artifacts in this conventional CT image. This is in agreement with our observation from Figure 20, where the SSOC time series had already predicted that the amplitude of the patient's heart movement was very small. This is confirmed also by the fact that the patient's heart size was very large, suggesting small heart motion amplitude. In this image (Figure 21) the characteristics of diagnostic importance are white dots that represent cross-sections of blood vessels that are full with calcification deposits. Blood vessel calcification is evident in this image by the two distorted white dots at the top right area of the heart.

The image in Figure 22 corresponds to the synthesized sinogram output of the SSOC and CSS processes. Although, this image provides better and brighter definition, compared to the conventional CT image, for the two white dots indicating the cardiac blood vessel calcification at the top right area of the heart, the overall quality of the corrected image is not better than the conventional CT image. This is because the cardiac motion artifacts in this case were not very strong and the base image interpolation process has not been very effective as this is indicated by artifacts shown as white stripes across the corrected image, which is a similar case in terms of performance with the image in Figure 19 for human patient#1.

Figure 23 shows the corrected image for human patient#2 using the retrospective RG-SSOC gating method on the same set of data of Figure 20. The starting point for the retrospective RG-SSOC process was the projection point#3988 with the SSOC output waveform indicating low motion without gasping. The quality of motion correction with the RG-SSOC method was consistent at various projection points indicating low motion by the output of the SSOC. In this case, the motion correction results of the RG-SSOC method are superior to the CSS method shown in Figure 22.

Thus, the conclusions drawn from this data set are the following.

- When the SSOC time series have very small amplitude for organ motion effects then there is no need to apply motion correction as defined by the CSS process, since the conventional CT images have sufficient clarity and do not suffer from cardiac motion artifacts.
- The breath holding restriction should not be used when the SSOC and the RG-SSOC processes are being used to correct for motion artifacts.

The validity of both these two observations is examined next with a CT data set from the human patient#3.

V.2.3 Human Patient #3: 72 bpm cardiac motion

The data acquisition arrangements for the CT data sets of the human patient#3 were identical to those of the human patients #1 and #2 except for the CT period of rotation, which was set to 1.0 second. This change in the CT rotation period was made because the patient's#2 pulse rate was approximately 72 beats per minute. During the CT data acquisition process the patient was asked to breath freely.

The motion time series extracted by the software spatial overlap correlator are shown in the upper curve of Figure 24. The same time series after smoothing (low pass filtering) is shown in the second curve from the top. The properties of the heart motion of the patient are immediately obvious in these curves. In fact it is possible to compute the actual period of the motion from the motion curves, which is approximately 0.833 second. Moreover, the SSOC time series reveal that the amplitude of the patient's heart movement is constant during the CT data acquisition process and superimposed on a very low frequency motion, most likely due to breathing effects . This is an indication that when the patient is not restricted to follow breath-holding procedures then is more relaxed allowing for a better tracking of the heart movement by the SSOC process.

The time varying correlation function for one selected phase of interest is shown in the third curve from the top of Figure 24. The vertical bars in this curve indicate the local maxima of the time varying function. The vertical bars in the bottom curve show the projections that are selected as the center of segments for

coherent sinogram synthesis. The base image interpolation scheme was used to coherently synthesize a sinogram for the selected phase of the heart movement.

The image in Figure 25 shows the conventional x-ray CT image for subject #3. The tomography image of the heart is at the centre. The lung structure characteristics are also shown at both sides of the heart image with a tumor at the top left side of the heart. There are two kinds of motion artifacts in this conventional CT image. First, artifacts that are due to breathing effects, which are evident at the area near the sternum. Secondly, the cardiac motion artifacts, which are indicated by the white distorted dot near the top right area of the heart, which suggests blood vessel calcification.

The image in Figure 26 corresponds to the synthesized sinogram output of the SSOC and CSS processes. In this case, motion artifacts due to breathing effects have been removed as indicated by the clarity of the image near the area of the sternum. This was achieved by the CSS process that provided selection of sinogram segments representing the same phases of the heart and breathing motion cycles, as shown by the time series of Figure 25. Since the period of the breathing motion is long (2-3 seconds) compared to that of the heart's periodic motion (0.5 – 1 second), then another way to remove the breathing motion effects is to apply a band pass filter on the SSOC time series of Figure 26. Our experience suggests that the best approach to remove the breathing motion artifacts is to use the CSS process as shown by the results of Figure 26.

Another kind of improvement of diagnostic importance, in this case, is the better and brighter definition compared to the conventional CT image of the white dot indicating the cardiac blood vessel calcification at the top right area of the heart. Thus, the overall quality of the image in Figure 26 is significantly better than the conventional CT image. This is because the cardiac motion artifacts in this case were strong and the base image interpolation process was more very effective, compared to the previous cases of patients #1 and #2, to correct for artifacts when they exist.

Figure 27 shows the corrected image for human patient#3 using the RG-SSOC gating method on the same set of data of Figure 24. The starting point of the retrospective RG-SSOC gating was at the projection point#1454 with the SSOC output waveform indicating very low motion amplitude. The quality of motion correction with the RG-SSOC method was consistent at various projection points indicating low motion by the output of the SSOC. A comparison between the corrected images of Figures 26 and 26 regarding the performance characteristics of the CSS-SSOC and RG-SSOC methods, suggests that both methods have been equally effective in this case to remove motion artifacts.

To strengthen the consistency and predictability in performance of the retrospective RG-SSOC gating method, Figure 28 shows the reconstructed image for human patient#3 using the RG-SSOC gating method on the same set of data of Figure 24. The starting point of RG-SSOC gating was at the projection point#11570 with the SSOC output waveform indicating large amplitude motion effects. In this case, deterioration in the quality of motion correction indicates the predictability of the RG-SSOC method to select good or bad projection points associated with small or large motion amplitude shown by the output of the SSOC.

VI. CONCLUSION

The paper discusses a novel approach to tracking and removing organ motion artifacts in CT systems. The method first uses the spatial overlap correlator to sample information associated with the organ motion present during the conventional CT data acquisition process. A software implementation of the spatial overlap correlator (SSOC), a retrospective gating (RG-SSOC) or a coherent sinogram synthesis (CSS-SSOC) process that both are based on the output of the SSOC method, are then used to correct for motion artifacts by selecting sinogram segments representing the same phase of the motion cycles. Real experiments with moving phantoms, clinical tests with human patients and an operational CT scanner demonstrate the efficiency of the SSOC process to tracking the motion effects localized in the tomography

area of the CT slice. However, only the retrospective RG-SSOC gating method has been consistently successful to correct for cardiac organ motion artifacts. In fact, the clinical test results of this investigation with human patients indicate that the performance of the retrospective RG-SSOC gating method is more predictable than the established retrospective ECG gating method. This difference is due to the fact that the SSOC process tracks the motion effects localized in the tomography area of the CT slice, something that an ECG system is not designed to track.

Although the acquisition process of this investigation included continuous projection data during 21 rotations, the clinical implementation of the new retrospective RG-SSOC gating scheme would require continuous projection data from 1.5 to 2 rotations only. This would allow the spatial overlap correlator (SOC output) to track and differentiate the small motion amplitude phase from the large cardiac motion amplitude corresponding to the QRS cycle of an ECG waveform, shown in Figure 12. Furthermore, when the sinogram's projection point is identified by the SSOC process with phase corresponding to the smallest motion amplitude then a segment-sinogram is synthesized with 180 degrees plus the fan angle temporal length for image reconstruction.

Thus, the output of the SSOC process forms the basis of a retrospective RG-SSOC gating process, with minimum exposure time (1.5 to 2 rotations per slice), which suggests that the new gating method will be very effective for 3-D motion correction for multi-slice CT scanners. Furthermore, the fact that the retrospective RG-SSOC gating process can perform well in cases when the patient breaths freely, suggests the kind of improvements that can be made for the 3D CT acquisition process. This is because, the clinical results (outputs of the SSOC process, Figures 13 and 20) of this investigation suggest that it is better to ask the patient to breath freely during the CT acquisition process, in order to avoid the gasping motion effects that become dominant during the last 10-seconds of the 3D CT acquisition process.

Finally, these clinical results with human patients have reinforced the results of previous investigations [14-16] that processing schemes correcting for cardiac motion artifacts work best when they account for the diastole phase, when the heart is moving the least.

Thus, the main difference between the retrospective RG-SSOC gating method and the CSS-SSOC coherent synthesis method of this investigation is based on the fact that the CSS method fails to exploit the properties of the heart's diastole phase. However, the CSS method could provide superior results for cases that there is continuous motion, which is reproducible.

Furthermore, to minimize CT radiation exposure associated with the CSS process, an internal triggering process is suggested for CT medical imaging applications [12]. However, since this CT internal triggering is based on estimates of the actual tracking of the organ motion provided by the SSOC process, the new triggering approach should have the advantage that the whole triggering process may be averted if the organ motion is very weak. In other words, the SSOC process has the capability to assess whether the amplitude of the organ motion is very small or large. Thus, if the organ motion amplitude provided by the SSOC process is small, then a CT image will not suffer from motion artifacts. Therefore there is no need to use any motion correction process.

Preliminary assessment of the retrospective RG-SSOC gating processes with real results from human patients suggests that the approach of this investigation can be used to screen aging population for cardiac diseases with emphasis to detecting cardiac blood vessel calcification.

Finally, the results of our investigation suggest that an optimum solution to the overall motion correction problem is the adaptive processing scheme [12,13] that removes, as an adaptive noise cancellation process, the spectral characteristics of the motion from the sinograms.

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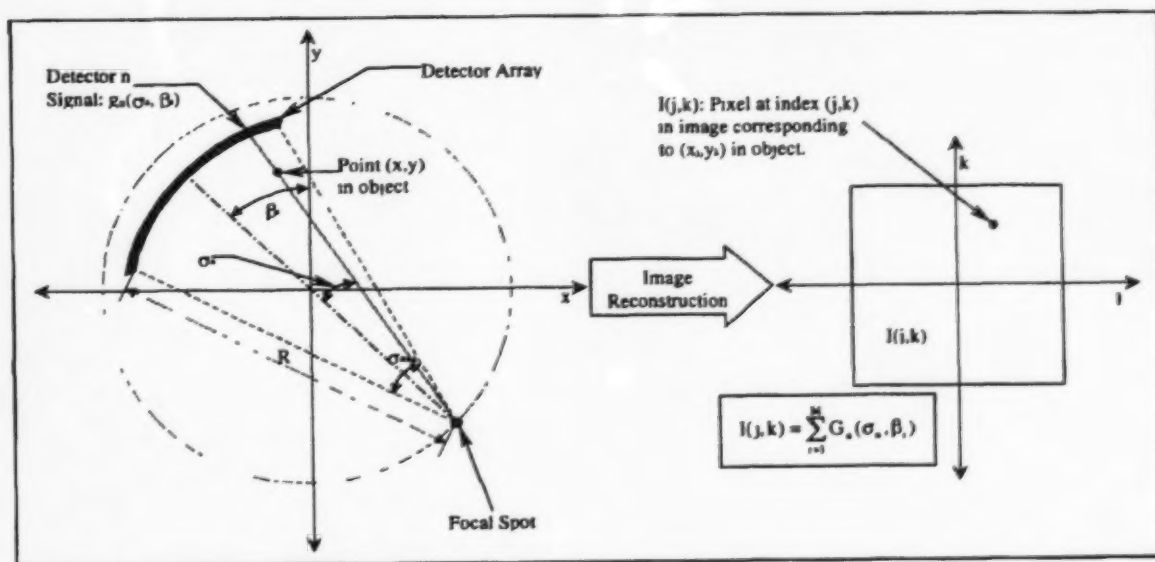


Figure 1: Schematic diagram of projection function for CT X-ray imaging systems.

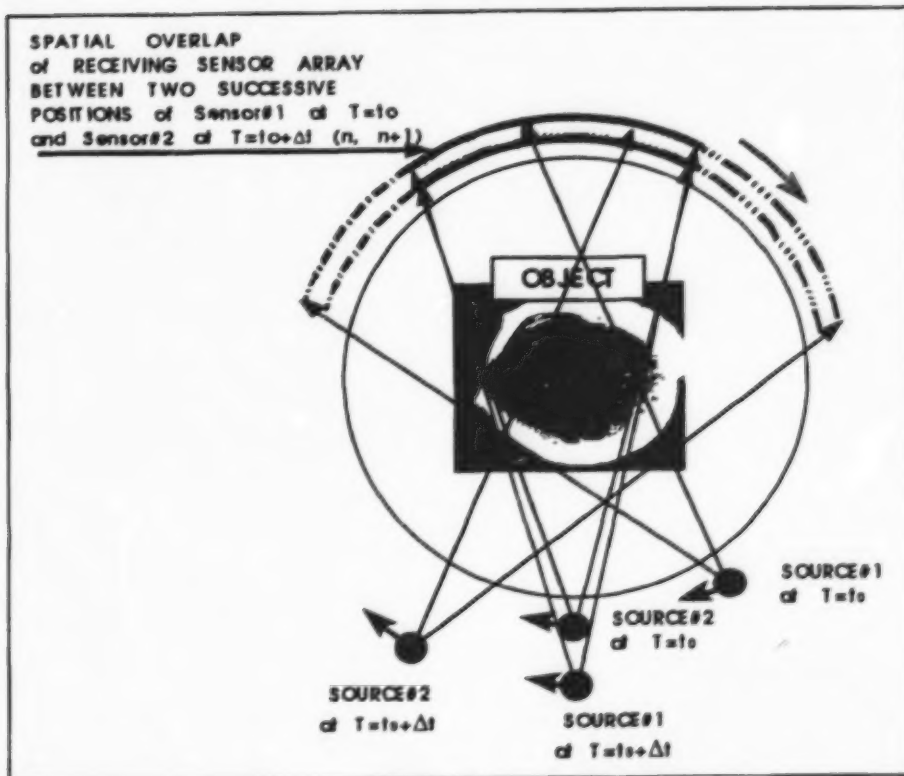


Figure 2: Two-source concept of spatial overlap correlator for CT X-ray imaging systems.

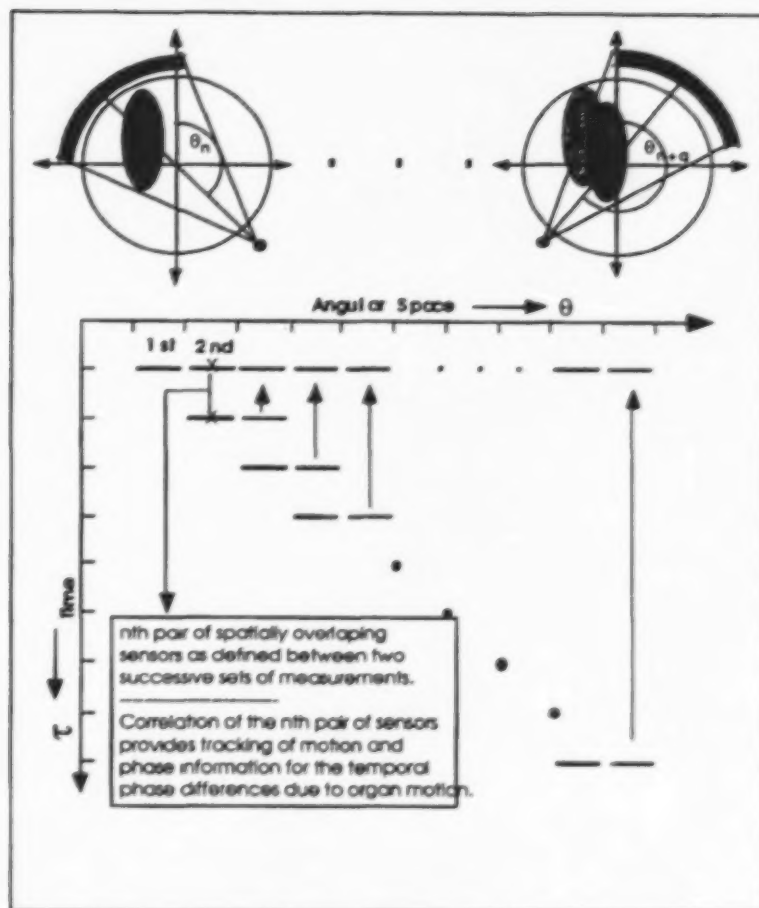


Figure 3. Graphical representation of the space-time sampling process of the spatial overlap correlator for x-ray CT imaging systems.

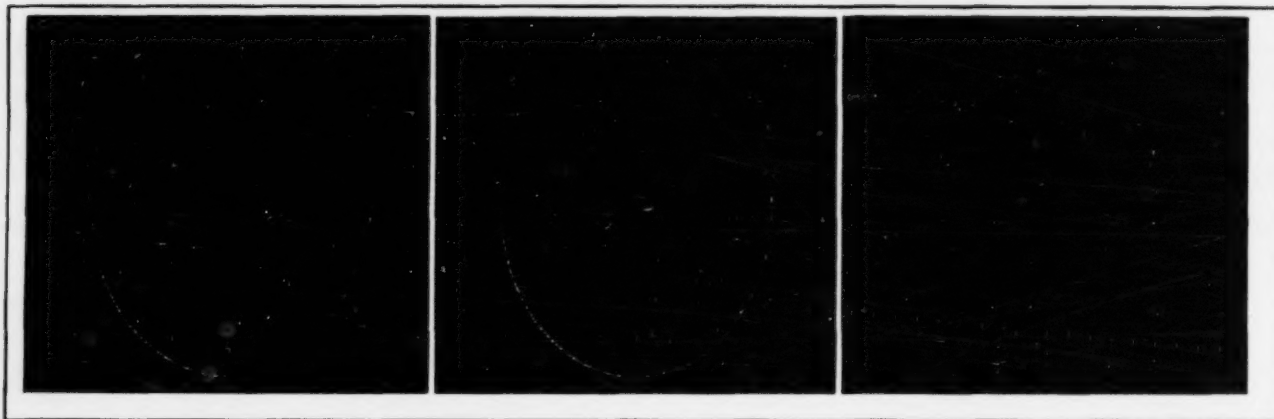


Figure 4: Simulation results for spatial overlap correlator. Left image shows the Shepp-Logan phantom. Center image includes motion at the left ellipse, which is taking place during the data acquisition period of the CT scanner. Right image shows the image reconstruction output of the spatial overlap correlator. None of the stationary components of the phantom appear in this image.

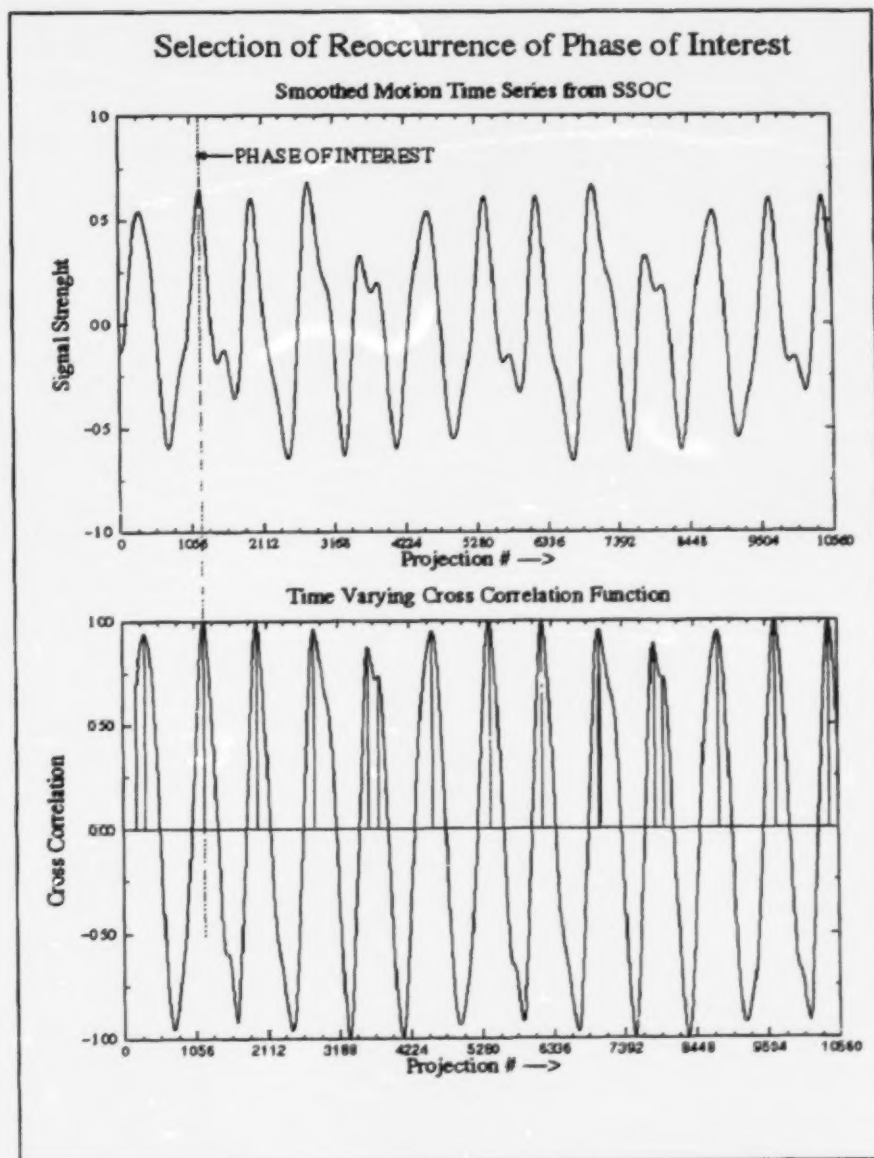


Figure 5 Selection of Phase of interest by correlation.

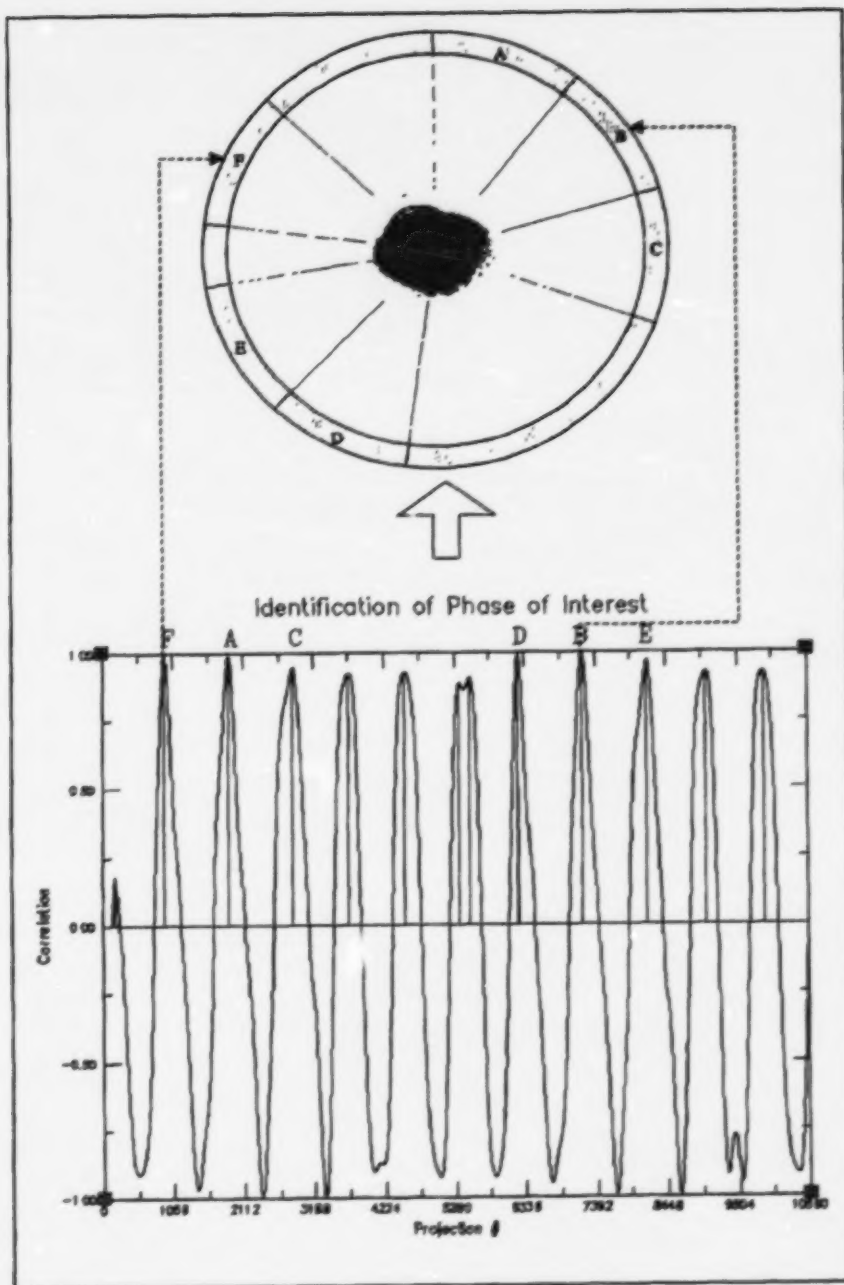


Figure 6 Operation of the Coherent Sinogram Synthesis Method.

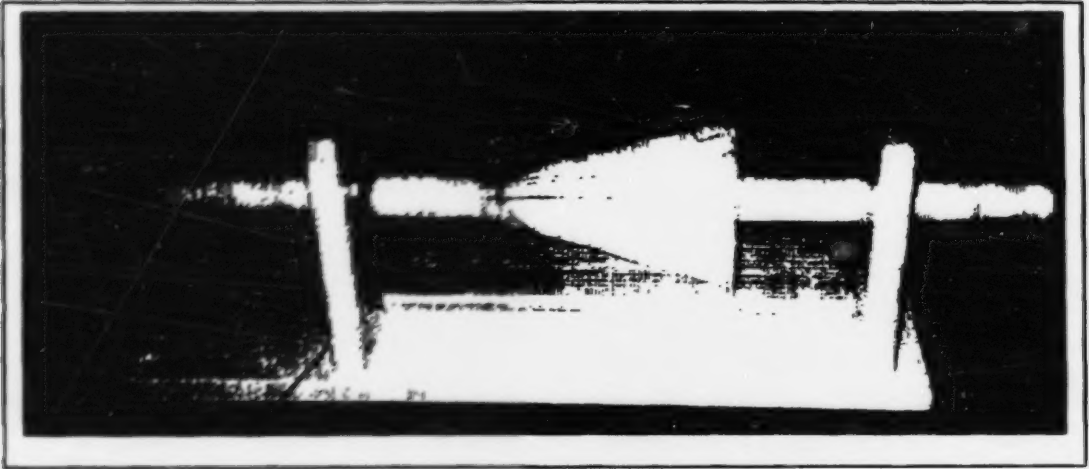


Figure 7 Close up view of Experimental Phantom.

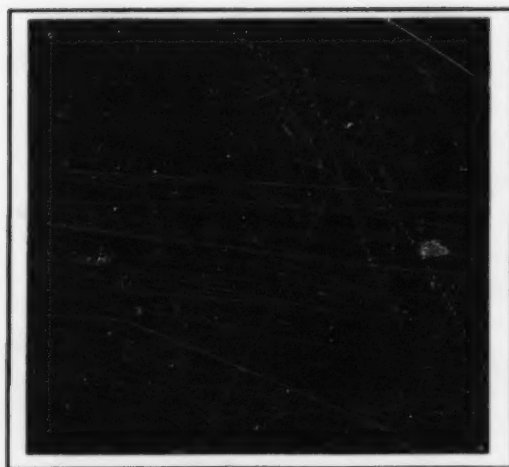


Figure 8 Reconstructed image by CT scanner for the stationary phantom, shown in Figure 7.

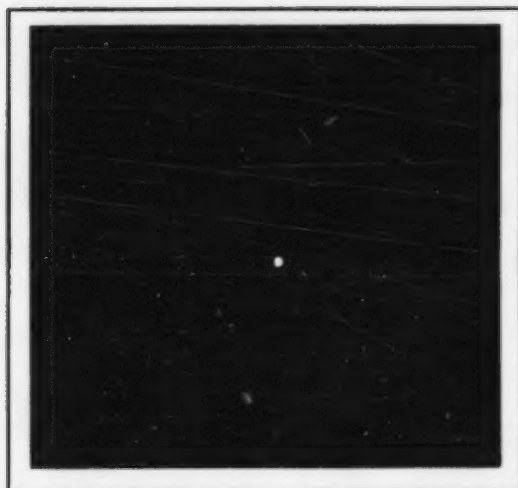


Figure 9 Reconstructed image by CT scanner for the moving phantom with period 0.6 second.

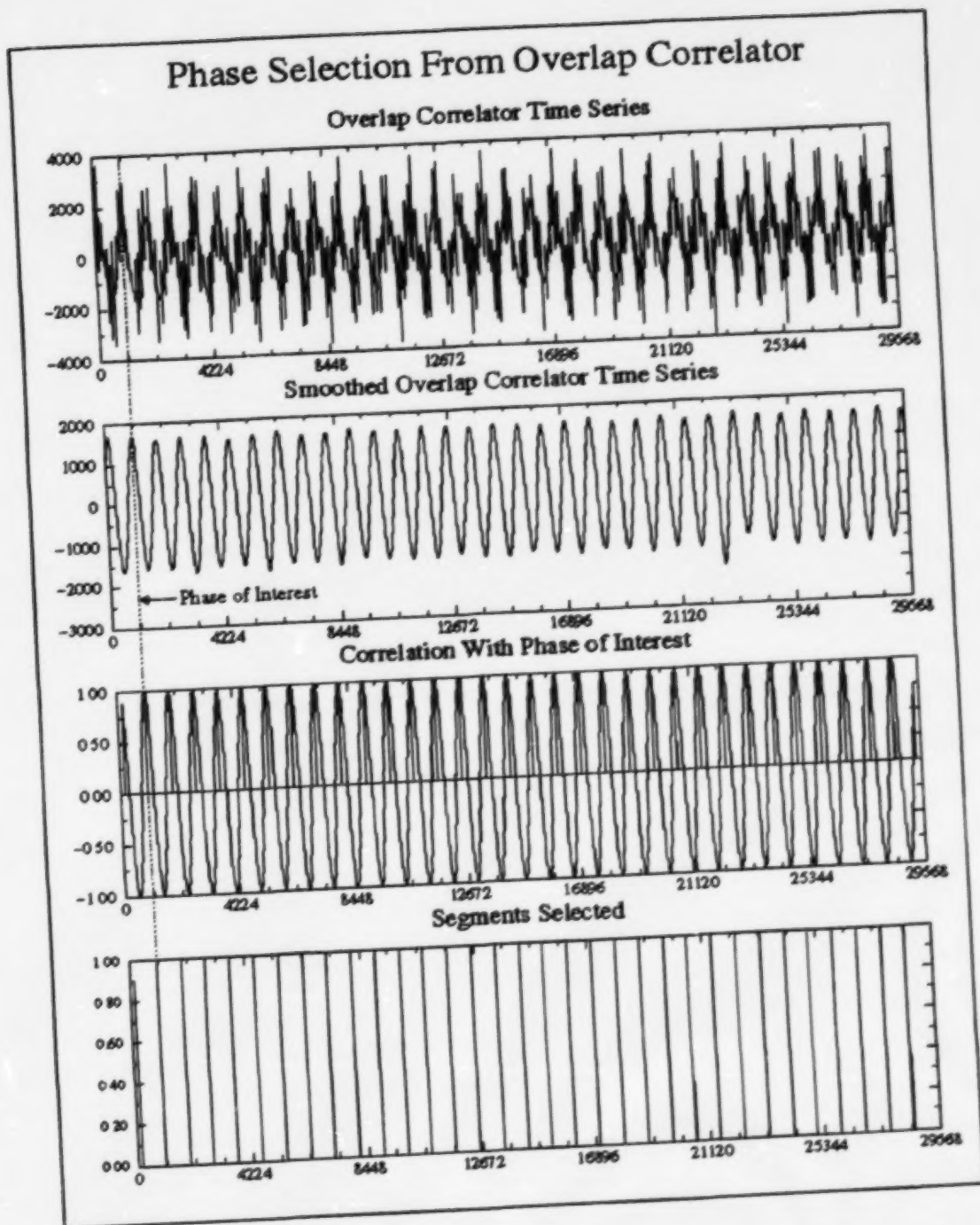


Figure 10: Time series tracking organ motion from Software Spatial Overlap Correlator for phantom motion period of 0.6 second.

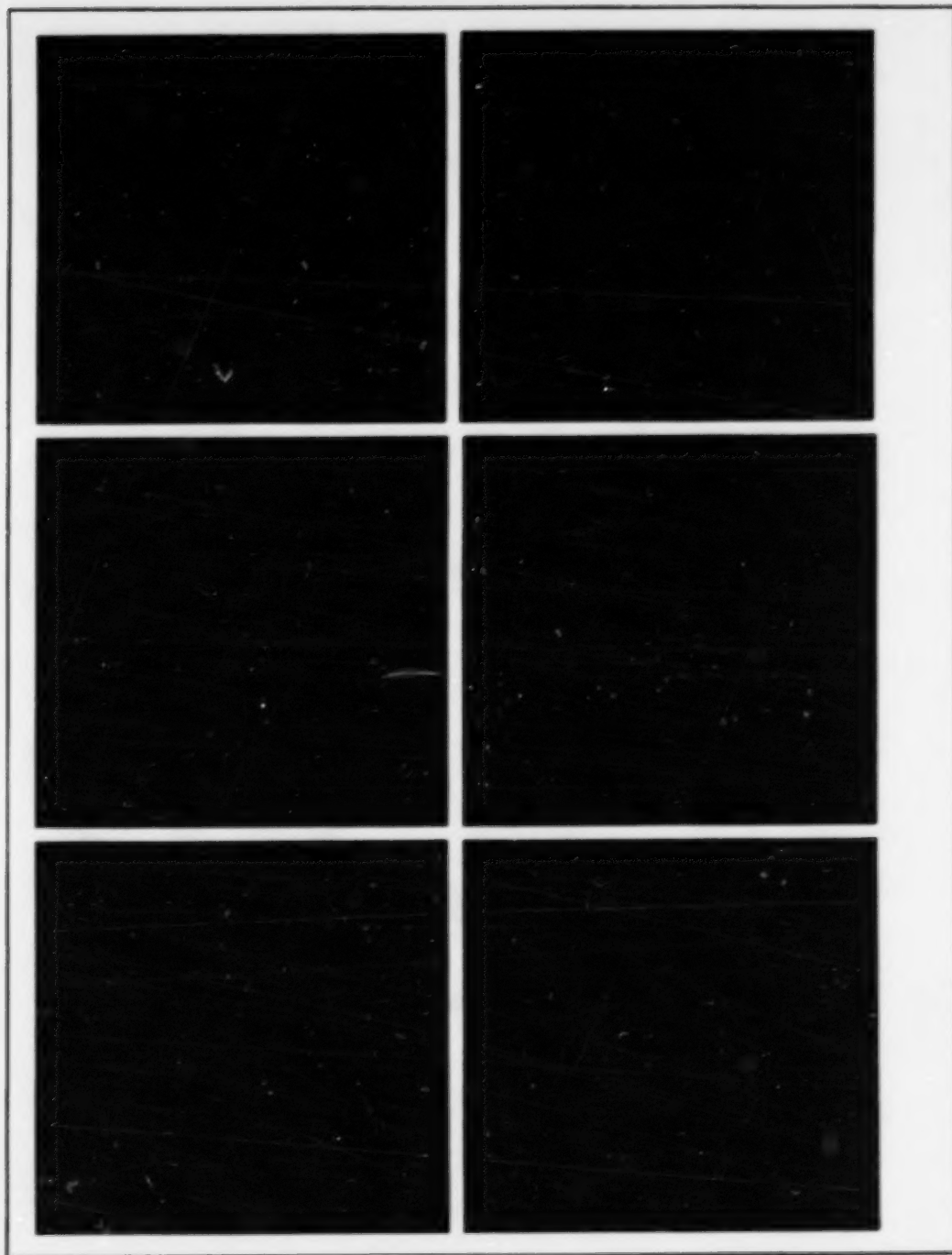


Figure 11: Various phases of resulting images using the CSS method with the SSOC with moving phantom with motion cycle period of 0.6 s.

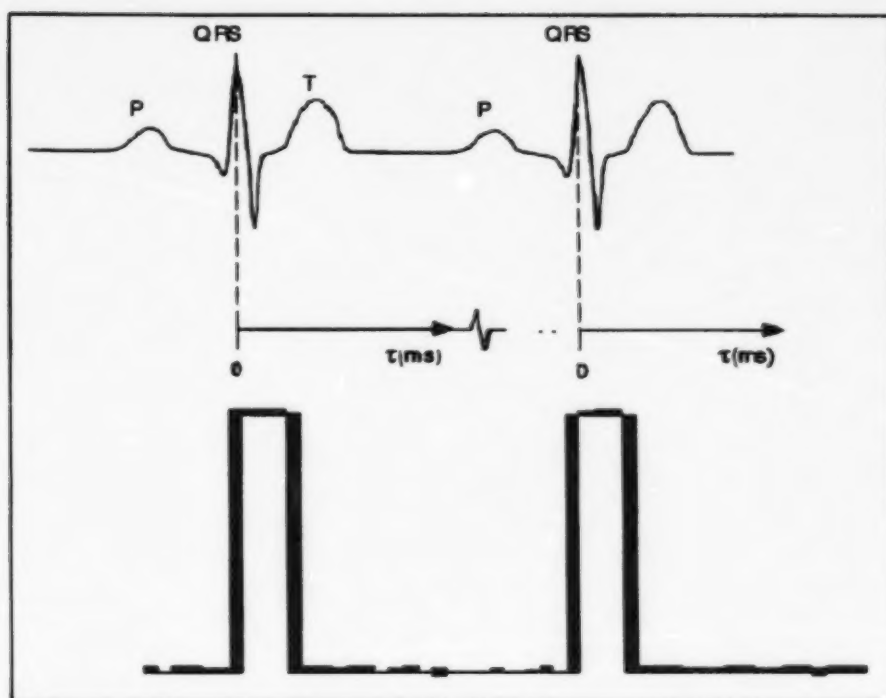


Figure 12 Synchronization process of ECG waveforms with recording of projection data from x-ray CT scanner. Upper Figure shows an ECG waveform and lower Figure the triggering signal from the ECG system to an x-ray CT scanner signaling the start of the x-ray CT acquisition process of projection data.

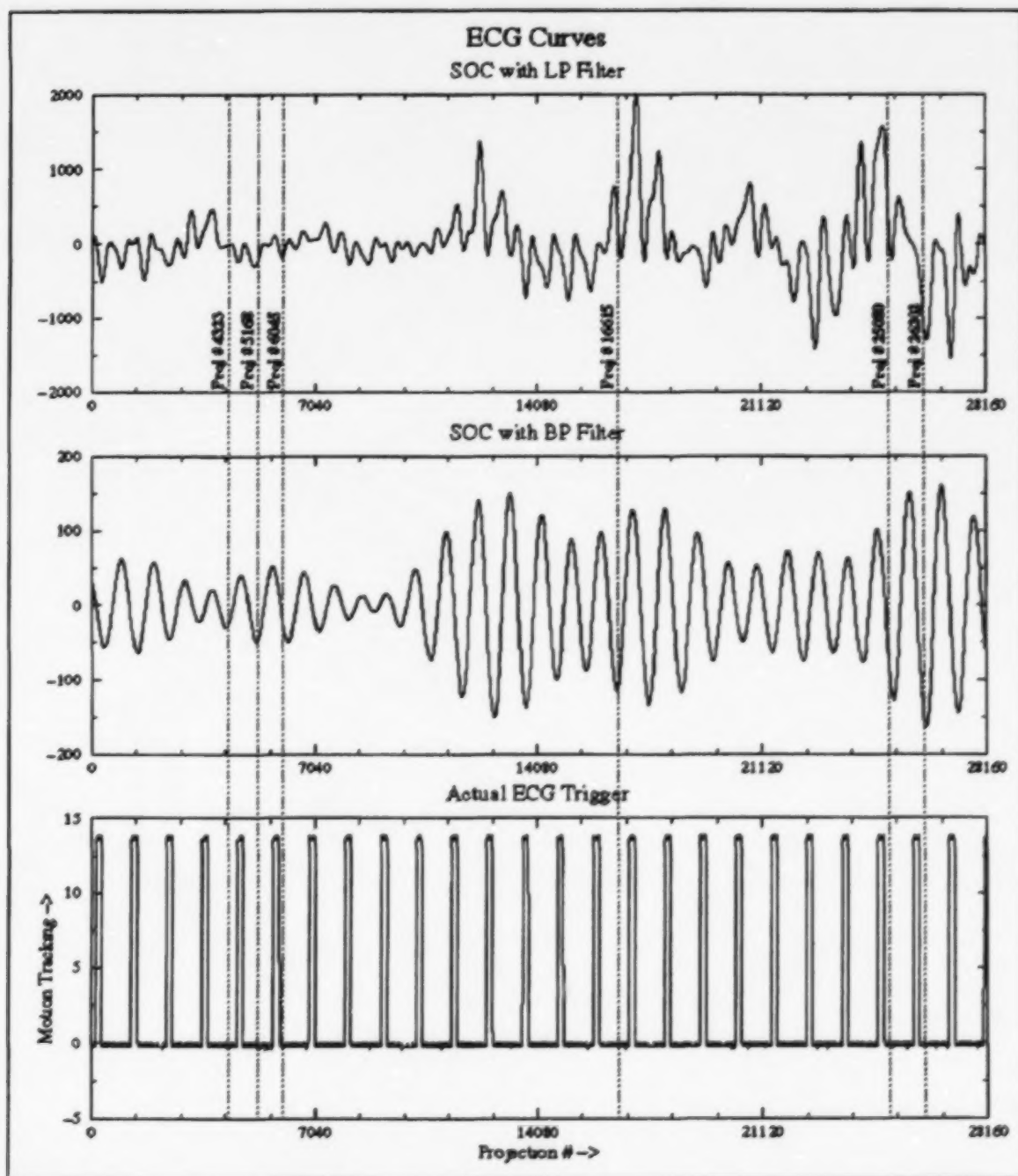


Figure 13 Time series (waveforms) from the output of the SSOC and the ECG triggering signal that allowed synchronization of the retrospective ECG gating with the RG-SSOC process. Upper waveform shows output of the SSOC. Middle waveform shows band-pass filtered output of SSOC. Lower waveform shows the ECG triggering signals synchronized with the starting point of the x-ray CT acquisition process.

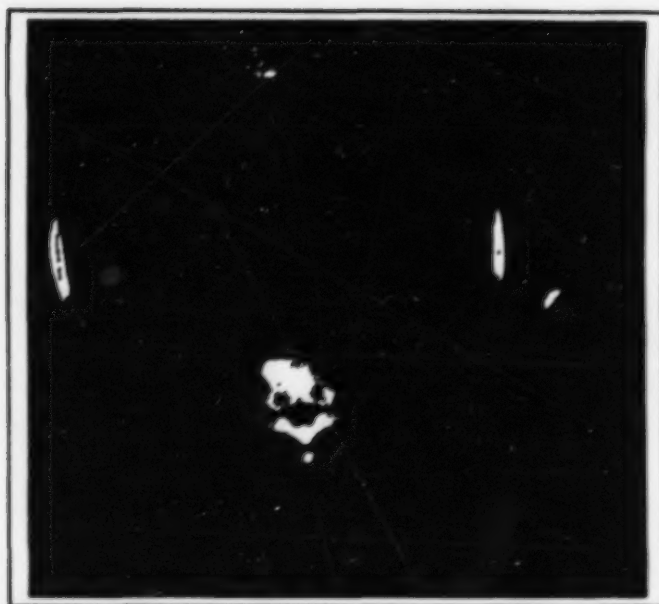


Figure 14 Output of conventional CT scanner for the human patient #1, with breath holding. The tomography area shows the lower part of the heart's ventricles.



Figure 15 Reconstructed image using retrospective ECG-gating at the projection point #6045, shown in Figure 13. This projection point defines the range between the T and P phases of the ECG-waveform and the starting point of the sinogram-segment with temporal length of 180 degrees plus the fan angle. The same projection point #6045 defines the area of the SSOC waveform (upper curve in Figure 13) with the least cardiac motion. Minimization of motion effects in this case allows identification of the heart's pericardium and its thickness, at the upper part of the image. The circular lobe at the left part of the image shows the tomography of the top part of the liver.

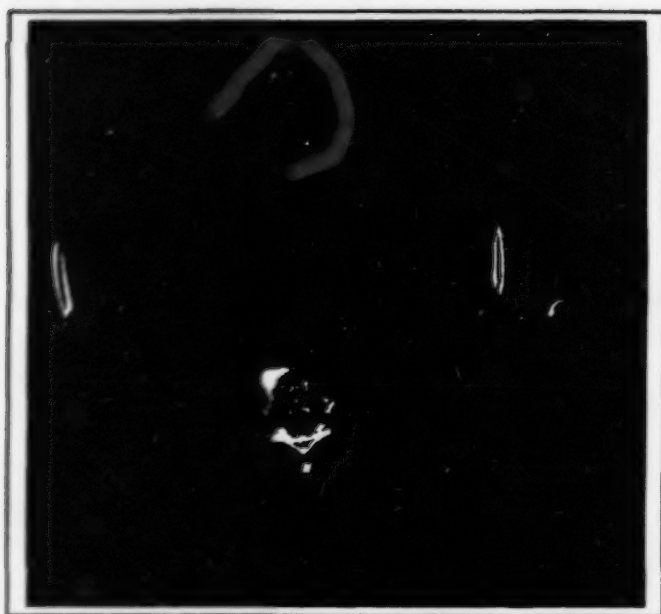


Figure 16 Reconstructed image using retrospective ECG-gating at the projection point #26202, shown in Figure 13. This projection point defines the range between the T and P phases of the ECG-waveform and the starting point of the sinogram-segment with temporal length of 180 degrees plus the fan angle. For the same projection point #26202, however, the SSOC waveform (upper curve in Figure 13) indicates large motion due to patient's gasping. Motion effects in this case are more prominent than those in the previous image in Figure 15.

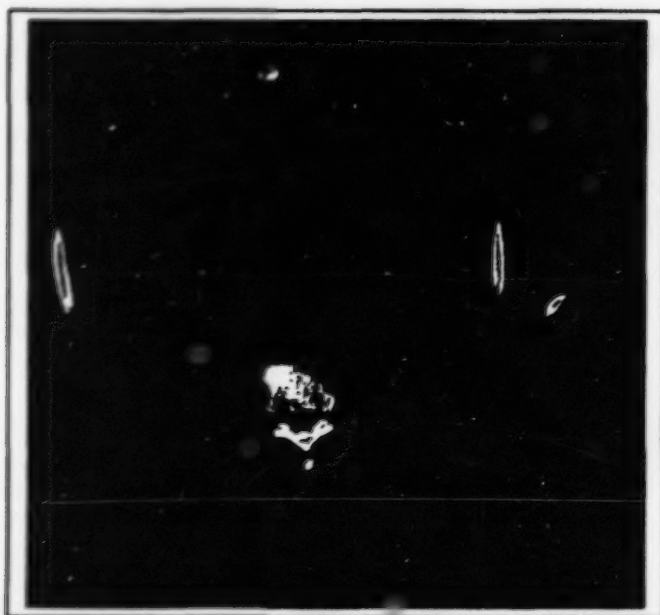


Figure 17 Reconstructed image using retrospective RG-SSOC gating at the projection point #4323, shown in Figure 13. For the same projection point, the ECG signal (lower part of Figure 13) suggests that the QRS phase of the ECG-waveform has not been excluded, which suggests the strongest heart motion effects. Despite the contradiction between the SSOC and ECH waveforms, motion effects in this image have been minimized as in Figure 16. In fact the thickness of the heart's pericardium as well as the boundaries of the ventricles can be identified with the same or better clarity as in Figure 15.

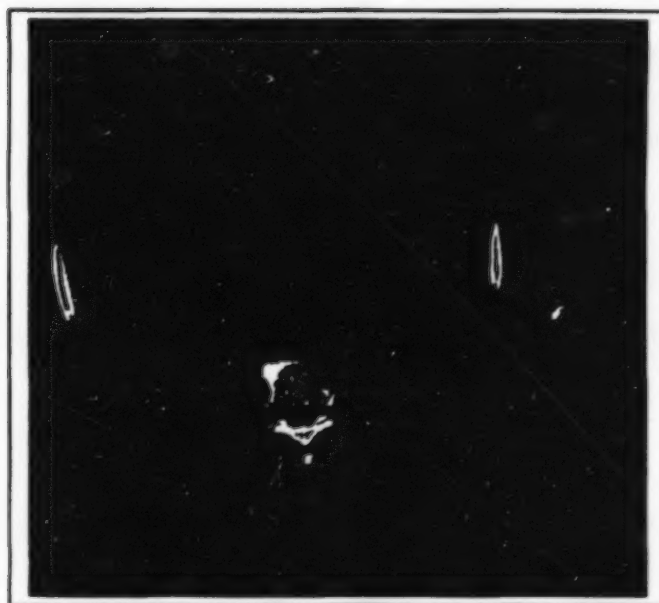


Figure 18 Reconstructed image from retrospective RG-SSOC and ECG gating from a set of projection data for human patient #1 with free breathing and with contrasting agent. The starting projection point for gating was chosen from the characteristics of the SSOC and ECG waveforms, both showing minimum motion, as in the case of projection#6045, in Figure 13. Motion effects in this case have been minimized as in Figure 16. The thickness of the heart's pericardium as well as the boundaries of the ventricles can be identified with the same or better clarity as in Figures 15 and 17.

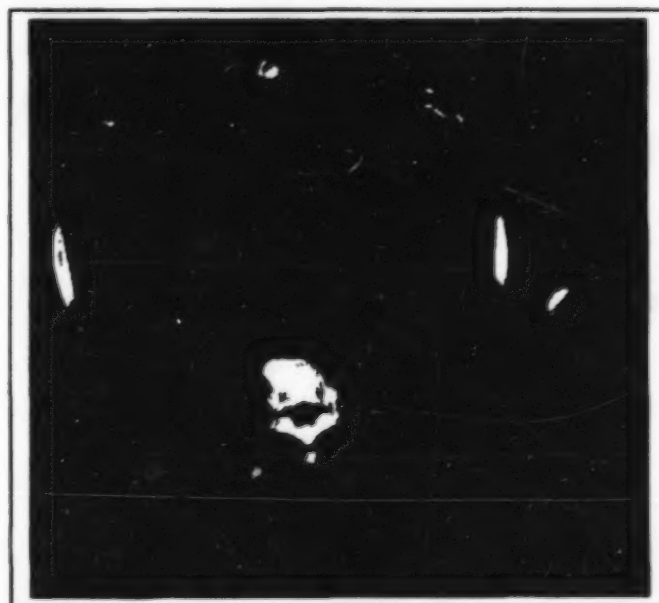


Figure 19 Corrected image for human patient#1 using the CSS-SSOC method on the same set of data shown in Figure 13. The corrected image in this case shows that the CSS-SSOC method has not been successful to minimize motion artifacts.

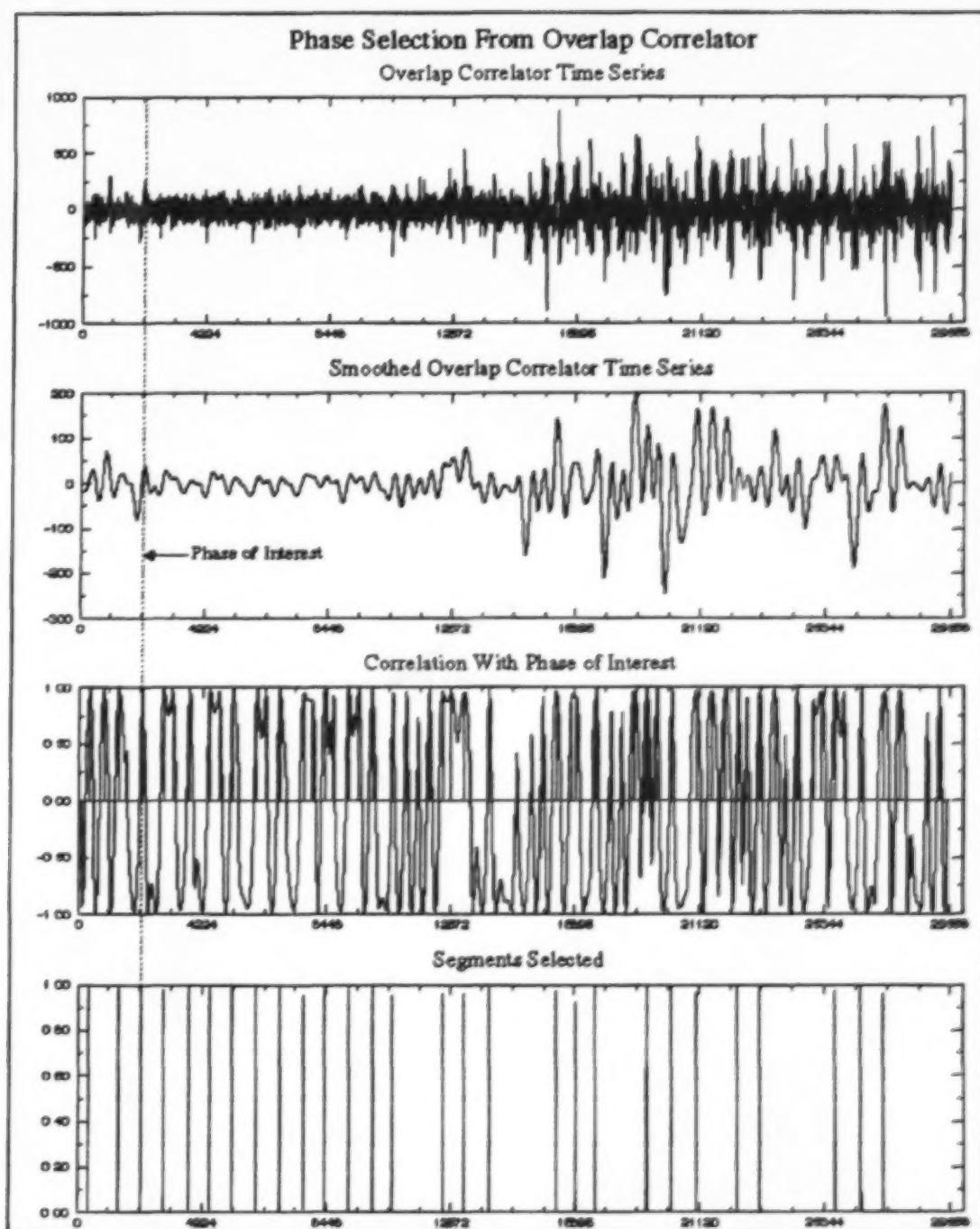


Figure 20 Time series (waveforms) tracking human patient's #2 heart motion from Software Spatial Overlap Correlator for motion period of 0.65 second. The patient was asked to follow breath holding procedure during the CT acquisition process.



Figure 21 Output of conventional CT scanner for a human patient#2 following breath holding.

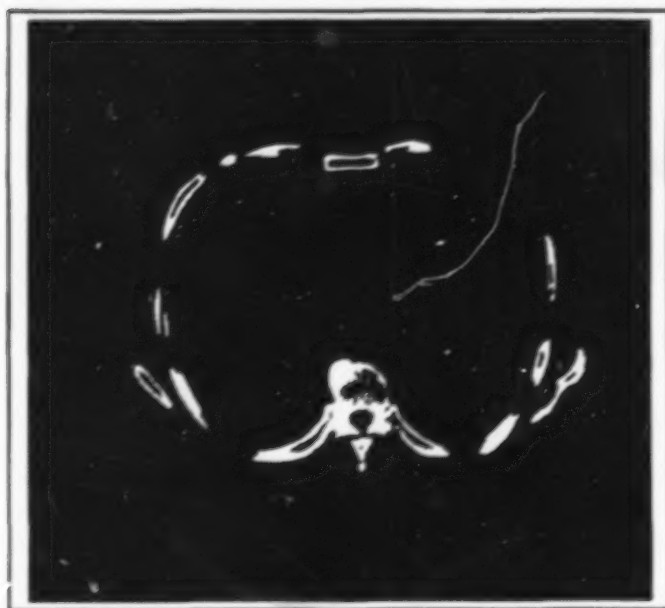


Figure 22 Corrected image for human patient#2 using the CSS method with the SSOC on the data sets of Figure 20.

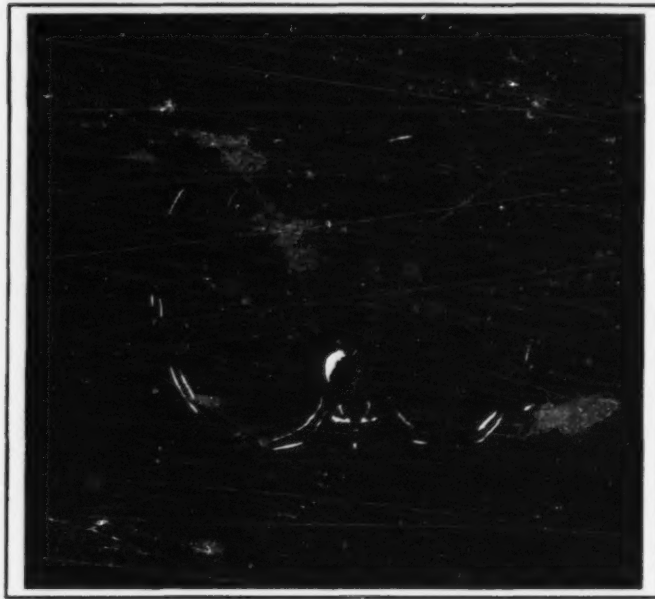


Figure 23 Corrected image for human patient#2 using the retrospective RG-SSOC gating method on the same set of data of Figure 20. The starting point of RG-SSOC gating was at the projection point#3988 with the SSOC output waveform indicating low motion without gasping. The high quality of motion correction provided by the RG-SSOC method was consistent at various projection points indicating low motion by the output of the SSOC. In this case, the motion correction results of the retrospective RG-SSOC gating method are superior to those of the CSS method shown in Figure 22.

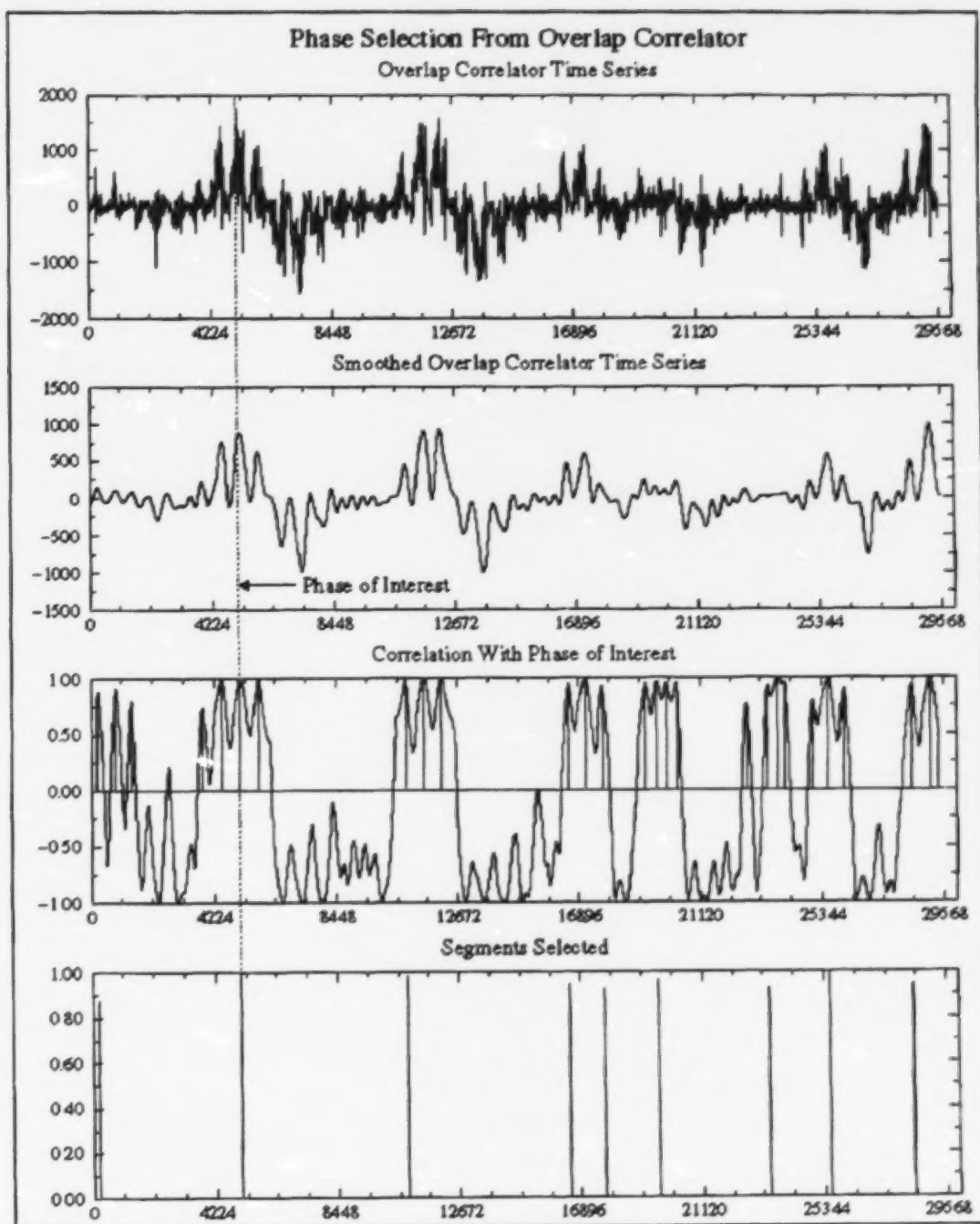


Figure 24 Time series (waveforms) tracking human patient's #3 heart motion from Software Spatial Overlap Correlator for motion period of 0.75 second. The patient was asked to follow breath free procedure during the CT acquisition process.



Figure 25 Output of conventional CT scanner for a human patient#3 following free-breath.



Figure 26 Corrected image for human patient#3 using the CSS-SSOC method on the data sets of Figure 24. The very good reproducibility of the motion effects, shown at the output of the SSOC in Figure 24, has improved the performance of the CSS method to correct the motion effects much better than in the previous two cases with the human patients #1 and #2.



Figure 27 Corrected image for human patient#3 using the retrospective RG-SSOC gating method on the same set of data of Figure 24. The starting point of RG-SSOC gating was at the projection point#1454 with the SSOC output waveform indicating very low motion amplitude. The quality of motion correction with the RG-SSOC method was consistent at various projection points indicating low motion by the output of the SSOC.



Figure 28 Corrected image for human patient#3 using the retrospective RG-SSOC gating method on the same set of data of Figure 24. The starting point of RG-SSOC gating was at the projection point#11570 with the SSOC output waveform indicating large amplitude motion effects. In this case, deterioration in the quality of motion correction indicates the predictability of the RG-SSOC method to select good or bad projection points associated with small or large motion amplitude shown by the output of the SSOC.